HOW DO WE ACCELERATE WHILE RUNNING?

by

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Running biomechanics are well established in terms of lower extremity joint kinetics as is the direct relationship between these variables and running speed. Many studies have investigated the differences in these variables when running velocity was increased in discrete increments but investigations of accelerated running in which velocity is continually increasing are almost non-existent. One investigation of the acceleration phase of running showed that joint torques did not increase while accelerating. These results cannot be aligned with the fully established results of running biomechanics at different speeds. We expected the joint torques to increase in magnitude for each step during the acceleration phase based on the previous research investigating increases in running velocity. The purpose of this study was to quantify lower extremity joint torques and powers during constant speed running and during running while accelerating at two rates of acceleration between a baseline velocity of 2.50 ms^{-1} to a maximal velocity of 6.00 ms^{-1}. It was hypothesized that lower extremity sagittal plane joint torques and joint powers would positively and linearly increase throughout the acceleration phase of running. 15 young, healthy runners (n = 8 females) between the ages of 18 and 22 were analyzed on an instrumented treadmill while accelerating at 0.40 ms^{-2} (A1) and 0.80 ms^{-2} (A2) from the initial to final velocities. Inverse dynamics were used to determine lower limb joint torques and powers using ground reaction forces and kinematic data collected by 3D motion capture. Correlation and regression analyses were used to identify the relationships between mean, maximum hip, knee,

and ankle torques and power to step number during the constant velocity and acceleration phase. The results of this study showed a significant increase in the joint torques and joint powers per step in both conditions A1 and A2 at the hip, knee, and ankle joints during the acceleration phase when the regression beta weights and correlation coefficients were tested for significance ($p < 0.05$). It was also observed that the knee and ankle joint torques and the hip, knee, and ankle joint powers had significantly greater increases per step in condition A2. There was no significant difference in the beta weights in hip joint torque between conditions A1 and A2. The constant state, pre- and post-acceleration phases had no relationship between joint torque and step number and joint power and step number in almost every variable, with three exceptions. There was a significant, direct increase in magnitude in hip joint power during the pre-acceleration period of condition A1, as well as hip joint torque during the post-acceleration period of condition A2. Additionally, a significant inverse relationship was seen in ankle joint power in condition A2 in the post-acceleration period. Finally, it was observed that the hip and ankle are the primary contributors to accelerating while running based on the magnitude of the beta weights of these variables, with the knee also contributing but not as much as the hip and ankle. In conclusion, in contrast to a previous study, our data suggest that hip, knee, and ankle torques doincrease during accelerated running on a step by step basis as do hip, knee, and ankle joint powers. Therefore, the tested hypothesis was supported based on the results of this study.

How do we accelerate while running?

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TABLE OF CONTENTS

LIST OF FIGURES

LIST OF TABLES

CHAPTER 1: INTRODUCTION

Basics of Running

Running is a repetitive, cyclic activity with a single cycle termed a stride. One stride cycle includes flight and support phases which are the periods of time the runner is not in contact with the ground and in contact with the ground, respectively. A stride is typically assessed from the initial heel contact of the foot to the next successive initial heel contact of the ipsalateral foot. Each stride is composed of left and right steps with a step referring to the initial contact of one foot to the initial contact of the contralateral foot (Thordarson, 1997). Running is distinguished from walking by one key difference. Running has a "flight phase" in which both feet are off the ground whereas walking does not (Dicharry, 2010; Nicola & Jewison, 2012; Thordarson, 1997). The support phase consists of an absorption or "braking" phase, followed by a propulsion phase. The swing phase is comprised of an initial and a terminal swing (Dicharry, 2010; Thordarson, 1997). During running there are forces acting on the body that tend to cause a collapse. The runner through lower extremity joint flexions and the anti-gravity, extensor muscles of the lower extremity work to prevent collapse during the entirety of the support phase (Winter, 1980). Winter (1980) stated that the lower limb support is derived from the combined muscle torques across the hip, knee, and ankle and are termed the "support" torque collectively. Support torques represents the ability to provide support and prevent collapse and is due to the collective activity of the muscles at all joints of the lower extremity. These muscles produce the individual joint torques and create the locomotive pattern of running. If there is a lack of torque production at one joint the other two joints may compensate for this to create sufficient support. This concept emphasizes the importance of examining all three joints when doing a kinetic assessment of gait (Winter, 1980).

The muscle groups at the three major joints of the lower extremity perform a combination of positive and negative work through their joint powers to create the running locomotion pattern. Power is calculated from the product of the torques produced at each joint and angular velocity (DeVita, Hortobagyi, & Barrier, 1998; Elftman, 1940; Johnson & Buckley, 2001; Winter, 1983). Positive power represents a concentric contraction of the musculature and negative power represents an eccentric contraction (Winter, 1980). These muscular contractions are what keep the body erect when running and prevents collapsing.

Background on Running Velocity Research

Running is involved in many sports and forms of physical activity. Within these activities running velocity is usually a vital factor to the athlete's performance. This has spurred research interest into running velocity and the changes in kinematics and kinetics that occur at different constant velocities. Research investigating the ground reaction forces with constant velocity finds the braking force impulse and the propulsive impulse to be equal in magnitude. Additionally it has been found that as velocity increases, the ground reaction forces- both horizontal and vertical- increase as well (Belli et al., 2002; Munro, Miller, & Fugelvand, 1987). Through the manipulation of running velocity, investigators have reported the relationship between velocity and joint torques and powers changes to have a direct relationship with increasing constant running velocity. The overall findings of the previous research indicated that as one runs at a faster constant velocity, the ankle joint torque will increase first contributing to the initial increase in velocity by means of increasing stride length, followed by an increase in joint torque at the hip and knee as velocity approaches maximal speeds by means of increasing stride rate (Belli et al., 2002; Dorn et al., 2012; Schache et al., 2011).

Background in Running Acceleration Research

The literature is quite limited in regards to the acceleration phase of running and the neuromuscular causes of actual acceleration. One novel study examined acceleration during running but the results do not seem to relate to the previous research that investigated increased constant velocities. Van Caekenberghe et al (2013) stated that the primary focus in running relating to varying velocities has investigated running locomotion during different constant velocities. Running, whether it be during a race or particular sporting event has periods of constant velocity but periods of acceleration are also involved, or a continuum of velocities involving purposeful acceleration phases. Van Caekenberghe et al (2013) found no significant changes in the joint torques in the acceleration phase when compared to the constant state velocity. However, they do suggest that power output by the muscles must be larger to increase speed and they found an increase in the positive power at the hip the negative power at the knee was greater. (Van Caekenberghe et al., 2013). This knowledge is verified by the previous literature that has investigated the changes in joint powers that occur during running (Cavagna & Kaneko, 1977). These findings do not seem to align with the findings of the studies that have shown when running velocity is increased, the joint torques also increase (Belli et al., 2002; Dorn et al., 2012; Schache et al., 2011). During the acceleration phase, one is running faster at a constant or variable rate of change thus the joint torques should increase during an acceleration phase. Another interesting finding of this research is that the antero-posterior ground reaction forces (GRF) on the treadmill were found to be equal in both the anterior and posterior direction. In order to run faster, the “propulsive” (anterior) reaction impulse needs to be greater than the "braking" (posterior) reaction impulse based on Newtonian mechanics (Hunter, Marshall, & McNair, 2005; Walter & Carrier, 2009). The findings of Van Caekenberghe et al (2013) are indicative of a constant velocity running period. They attribute these findings primarily to the lack of the body leaning forward on a treadmill and thus attribute them to the finding that the joint torques did not increase.

The findings of Van Caekenberghe et al (2013) are conflicting to the previous research investigating velocity modulation in constant states. Acceleration is an integral part of running performance and therefore it merits further investigation based on the minimal research that has been done on the acceleration period of running.

Hypothesis

Based on the previous research investigating running biomechanics, including velocity related changes in running biomechanics, it was hypothesized that lower extremity, sagittal plane joint torques and joint powers would positively and linearly increase throughout the acceleration phase of running.

Statement of Purpose

The purpose of this study was to quantify lower extremity joint torques and powers during constant speed running and during running while accelerating at two rates of acceleration (0.40 ms^{-2} and 0.80 ms^{-2}) between a baseline velocity of 2.50 ms^{-1} to 6.00 ms^{-1}.

Significance

The literature investigating the changes in running kinematics and kinetics at different speeds is much more extensive than the literature on the acceleration phase of running. The research on running at different speeds indicates greater joint torques and powers when one runs faster, which is in contrast to the findings of Van Caekenberghe et al (2013) who report there are no differences during the acceleration phase of running when compared to steady state running. This study was intended to cross-validate these results and in doing so adding to the literature on accelerated running. Additionally, this study was meant to add to the minimal literature that investigates the acceleration phase of running.

Delimitations

1. The subjects will be young, experienced runners who run at least 6 miles a week.
2. Subjects will be male and female between the ages of 18 and 25 and will have natural differences in gait kinematics.
3. The subjects will be running on a Bertec Instrumented Treadmill, which is an automated treadmill controlled by the researchers with a force plate embedded below the belt.
4. The only kinematic measurements will be taken at the hip, knee, and ankle on both legs.

Limitations

1. There is some error that may occur from soft tissue artifact or from the cameras that are used during motion capture.
2. As with the motion capture system, the force plates may not always accurately measure the magnitude or the location of the ground reaction force vector.
3. The previously mentioned limitations could lead to error in the inverse dynamics inputs of joint torques and joint powers.

Operational Definitions

1. Accelerated running- the period of time in which the participant is increasing their velocity from the baseline velocity to the maximum velocity.
2. Ground reaction force- the force that acts equal and opposite to the force created from the participant being in contact with the surface of the force plate.
3. Power absorption- the negative power created from the eccentric (muscle lengthening) contraction of the anti-gravity muscles.
4. Power generation- the positive power created from the concentric (muscle shortening) contraction of the anti-gravity muscles.
5. Posterior“Braking” force- the posterior component of the ground reaction force.
6. “Propulsive” force- the anterior component of the ground reaction force.
7. Joint torque- the muscular contributions at each joint, synonomous with joint moment.
8. Stride length- the distance from the ipsilateral foot to the ipsilateral foot
9. Step Length- the distance from the ipsilateral foot to the contralateral foot

CHAPTER 2: LITERATURE REVIEW

This study investigated and quantified any changes in joint torques and joint powers in the lower extremity that occur between constant speed running and the acceleration phase of running at two different rates of acceleration between a baseline velocity of 2.50 ms^{-1} and 6.00 ms^{-1}. A regression based approach was utilized to determine the relationships between these variables and the sequence of steps through the acceleration phase. This chapter will review the scientific literature related to running biomechanics and is portioned into these sections; Advanced Analysis of Running, Literature on Running at Different Constant Velocity Rates, and Literature in Accelerated Running.

Running is one of the most common forms of physical activity in the United States. As of 2008 the total number of runners, whether it be recreational or competitive was approximately 35,904,000 (Cooper, 2009). Interest in the analysis of running originates in the times of the Ancient Greeks who were fascinated by the artistic display of running; Aristotle, in particular had an interest in running locomotion and the differences seen in humans and in the animal kingdom. Wilhelm and Eduard Weber in 1836 truly initiated the study of gait in running and walking formulating over 150 hypotheses (Cavanagh, 1990). While distance running has always been a part of human locomotion, it was not particularly popular in terms of participation until the late 1960’s and early 1970’s in the United States (Cavanagh, 1990) and this increase in participation was at least partially due to the work of Kenneth Cooper and his book entitled “Run for Your Life: Aerobic Conditioning for Your Heart” (Houmard, 2013). Running is also involved in a variety of physical activities including soccer, football and basketball. Since running is an integral part of these activities running research contributes valuable information as how to improve performance.

Advanced Analysis of Running

Understanding the basic concepts of the kinematics and kinetics of running is essential for the purpose of this study. The gait cycle refers to the events that occur in the lower extremity during running between the initial contact of one foot to the next successive initial contact of the same foot which is called the stride (Thordarson, 1997). A step is the period between the initial contact of one foot to the initial contact of the opposite foot. Running is an extension of walking with one key difference; this difference being the addition of a “flight” phase and lack of a “double stance” phase (Dicharry, 2010; Nicola & Jewison, 2012; Thordarson, 1997). The “double stance” phase occurs in walking in which both feet are on the ground. The “flight” phase is the period during running in which both feet are not in contact with the ground (Dicharry, 2010; Thordarson, 1997).

The gait cycle can then be broken down into two sub-phases, a swing phase and a support phase. The support phase looks at the foot when it is in contact with the ground and is further broken down into an absorption phase and a propulsive phase. The swing phase is broken down into the initial swing and terminal swing (Munro et al., 1987; Thordarson, 1997). There is some variance among the research in terms of percentages of each phase, but the general consensus is that the swing phase occupies a greater percentage (approximately 60 percent) of one stride or cycle and the support phase is a lesser percentage (approximately 40 percent). This also will vary with running speed, where when one runs faster the swing phase percentage increases (Nicola & Jewison, 2012; Thordarson, 1997).

Elftman (1938) is one of the earliest accounts in the investigation of gait locomotion biomechanics. He performed a classical study that was able to quantify the movements of the joints during locomotion with the use of a three joint model (Elftman, 1938; Elftman, 1940). The ankle and knee joint have been found to play a dominant role in the support phase, providing higher magnitude joint torques at the absorption and propulsion phases and thus preventing collapse through the support phase (Arampatzis et al., 1999; Belli et al., 2002; Winter, 1983). During the swing phase, the hip flexors are concentrically contracting during initial swing followed by the hip extensors eccentrically contracting in the terminal swing in preparation for ground contact. The knee is also active during the swing phase with knee extension occurring in the initial swing. The ankle is mostly active during the stance phase producing plantarflexor torques (Winter, 1983). Elftman's (1938) early models are what allowed for such findings and continued regarding the kinematics of running. The patterns of joint mechanics at the hip, knee, and ankle are described further in the next paragraphs.

Everybody has a certain uniqueness in the way in which they run and these subtle differences in kinematics and kinetics can be seen in previous research quantifying muscle activation patterns. However, while the magnitudes in the EMG data may vary between people, a common pattern of muscle activation can be seen (Guidetti, Rivellini, & Figura, 1996). In terms of creating the locomotion pattern of running the three large joints in the lower extremity, hip, knee, and ankle, work as one unit with the muscles creating movement and support through the gait cycle (Winter, 1980). Winter (1980) derived the equation;

$$M_s = M_k - M_a - M_h$$

Where M_s is the net support torque of the three joints, M_k is the positive torque of the knee, M_a is the joint moment of the ankle, and M_h is the joint torque of the hip. The joint torques can also be seen by figure 1 below showing a visual representation of the lower extremity joint torques. In this picture counter-clockwise movements (knee) are positive and clockwise movements (hip and ankle) are negative. If one joint is not providing the normal, adequate support, the other joints in healthy individuals will be able to compensate for this inadequacy (Winter, 1980).

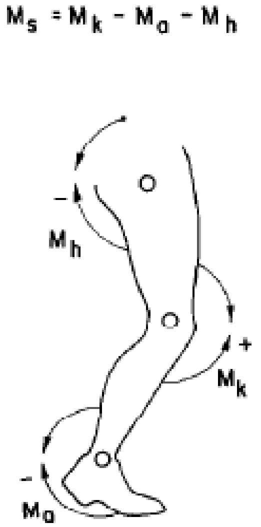

Figure 1- Moments (Torques) of force at the joints of the lower extremity (Winter, 1980)

The joint torques that occur during running can be quantitatively seen in figure 2 and the joint power patterns in figure 3 (Winter, 1983). The top curve represents the total support of the hip, knee, and ankle. Based on the figure from Winter (1983) the figures can be interpreted as follows:

During stance, we see that the hip in the initial stage is creating an extension torque through a very slight concentric contraction indicated by the positive power seen in figure 3. After midstance, there is a flexion torque corresponding to a negative power or eccentric contraction of the hip flexor musculature. Powers can be indirectly used to determine the type of contraction the muscles around a joint are performing. Upon entering the swing phase the hip flexion torque continues but it is occurring as result of concentric contraction. As the limb moves toward the terminal swing phase the hip extends concentrically to prepare the limb for the next contact.

The knee joint during stance creates an extension torque which is to prevent collapsing from occurring and is done so through eccentric contraction in the first half of stance and concentric contraction in the last half (figures 2 and 3). For this reason, the knee serves primarily as an “absorber” (Winter, 1983). During the first half of swing, the knee has an extension torque followed by a flexion torque created through the eccentric contraction of knee flexors to prepare the leg for contact.

The ankle joint has been determined to be primarily a “generator” (Winter, 1983). During the stance phase there is a plantarflexor torque created through an eccentric contraction of the ankle plantarflexors in the first half of stance acting as a shock absorbing mechanism; in the second half of stance the ankle plantarflexors start contracting concentrically to initiate toe-off. After toe-off and during the swing phase there is little occurring at the ankle joint as seen in figures 2 and 3 (Winter, 1983).

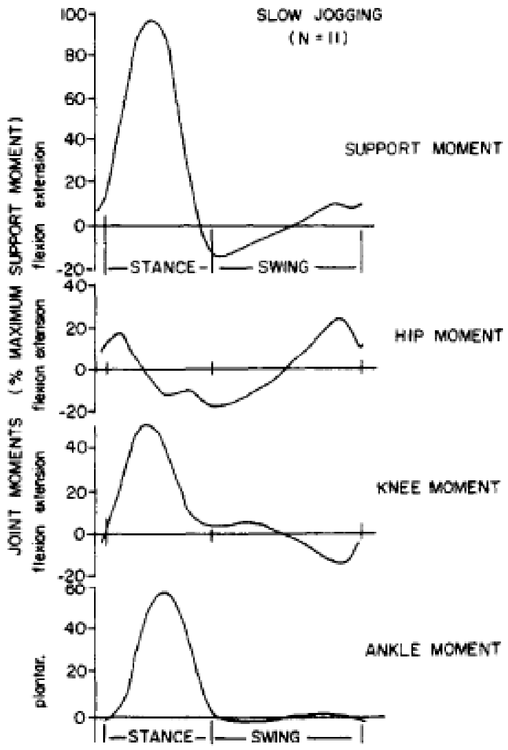

Figure 2- Joint torque patterns during running (Winter, 1983)

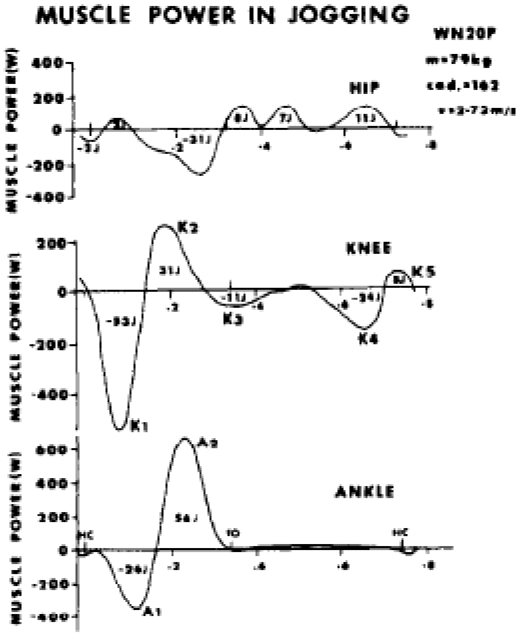

Figure 3: Muscle power patterns during running (Winter, 1983)

To summarize, the gait cycle is the period of events that occur throughout running locomotion. Running locomotion is supported and produced through the interaction of the hip, knee, and ankle joints. At initial contact, the ankle plantarflexors and knee extensors are eccentrically contracting to absorb the ground reaction forces. As the limb progresses through the support phase, a hip extensor torque occurs followed by a hip flexor torque through the end of support. The ankle then pushes off with a plantarflexor torque into the swing phase where the hip flexors contract concentrically bringing the limb forward. In the terminal swing, the hip extensors eccentrically contract to prepare the limb for the next ground contact.

Literature on Running at Different Constant Velocity Rates

As mentioned, running is involved in a variety of physical activities. These activities can serve as recreational forms of exercise or they can be involved in a competitive setting. In either environment, the velocity- directional speed at which we run- is crucial to the performance regardless of if it is a short sprint or a long distance run. Velocity is also rarely held constant and is more of a continuum than a discrete variable. Changing velocity has many purposes. It could be the end of the race and the participant could be trying for that last push to beat out an opponent across the finish line, or to get to the ball first in soccer. These would be examples of purposeful positive accelerations-increasing velocity over a certain time period. The opposite would be negative acceleration- a purposeful decrease in velocity. This may occur when a runner wants to decrease velocity to conserve energy for a long race.

The kinematics and kinetics of running at different constant velocities have largely been the focus of study in terms of changes in velocity (Van Caekenberghe et al., 2013). Studies such as the one performed by Dorn et al (2012) analyzed the muscular and joint aspects that change during four different running velocities, 3.5 ms^{-1}, 5.0 ms^{-1}, 7.0 ms^{-1}, and 8.0 ms^{-1} or greater (Dorn et al., 2012). The findings of this study stated that the initial increase in velocity, up to 7.0 ms^{-1}, is due to an increase in stride length (Figure 4). This has also been stated in previous literature that it is stride length first that increases velocity followed by stride rate at the higher velocities (Cavanagh & Kram, 1990; Fukunaga, Matsuo, Yuasa, Fujimatsu, & Asahina, 1980).

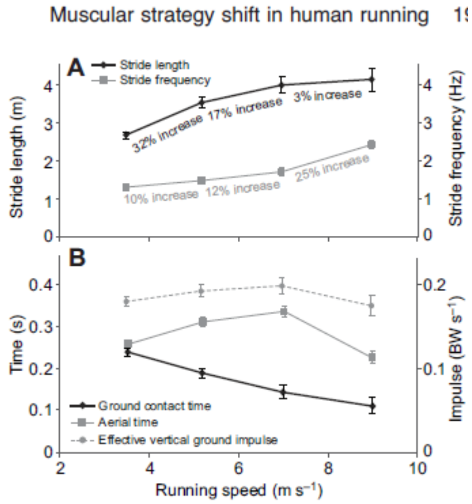

Figure 4. Stride length and stride frequency plotted against running speed. (Dorn et al., 2012)

They attribute this increase in stride length to the large joint torque created by the gastrocnemius and the soleus during propulsion which relates to the previous findings of other research projects (Hamner & Delp, 2012; Schache et al., 2011). The peak ankle torque occurs roughly 20 percent before the foot off instant, which can be seen in figure 5. Contributions to the vertical ground reaction forces remained roughly the same (Figure 6). In Hamner and Delp (2012) it was also found that with increased velocity the overall joint angles of the hip, knee, and ankle were greater which could have been a result from the increased torques produced by these joints.

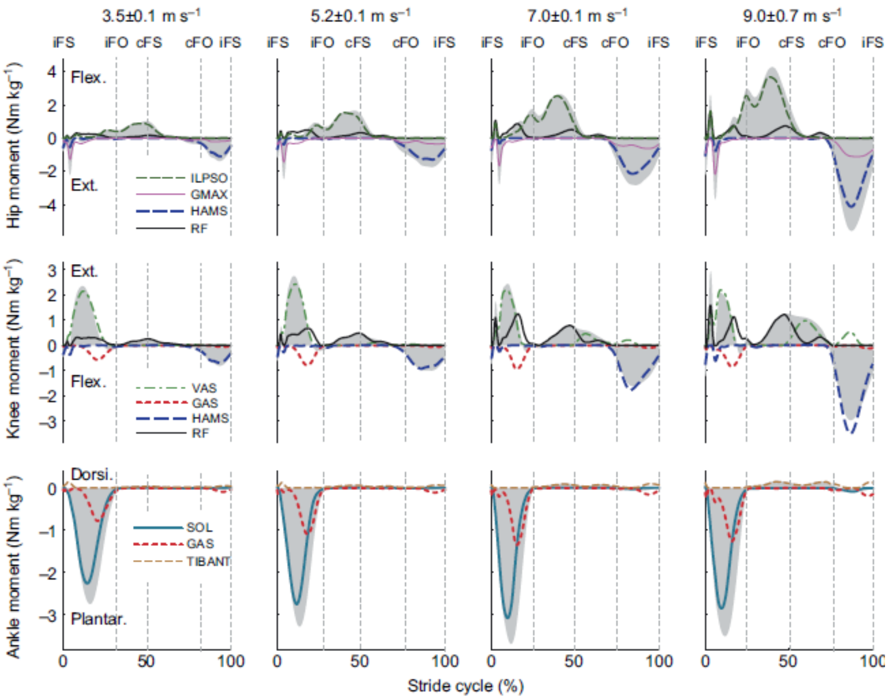

Figure 5 Individual muscle joint torques for each speed. (Dorn et al., 2012)

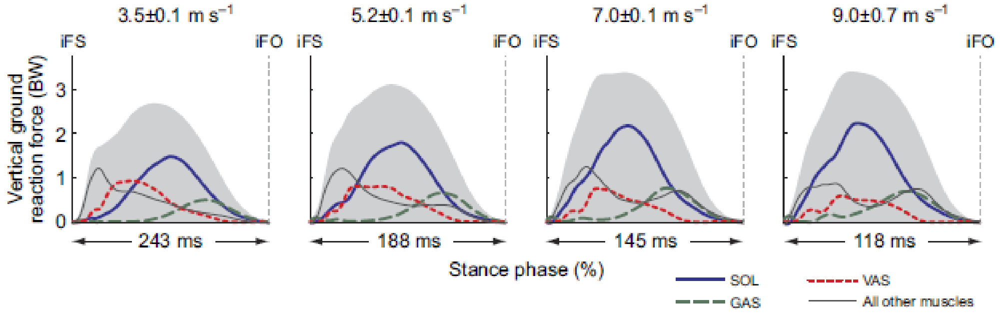

Figure 6. Net Vertical ground reaction forces (shaded). (Dorn et al., 2012)

Stride frequency continues to increase velocity once maximal stride length has been achieved, usually around 7.0 ms^{-1} (figure 4). The increase in stride frequency is driven primarily through the increased hip flexion joint torque that occurs in the swing phase. The iliopsoas drives hip flexion concentrically in the first half of the swing phase (figure 5). The hamstrings and the gluteus maximus create an extensor joint torque in the latter half of the swing to prepare for ground contact (figure 5). As one progresses to faster speeds, a larger joint moment can be seen in these muscles (figure 5) (Dorn et al., 2012). This finding aligns with the findings of another study that also found that a faster running speed is a result of a larger leg swing due to increased hip flexor and extensor action (Gazendam & Hof, 2007). These results also relate to the finding of Arampatzis et al (1999). In addition, they found that as running velocity is increased there is an increase in the leg stiffness driven by an increase in joint torques at the knee and ankle (Arampatzis et al., 1999). These data indicate that as one runs at a faster velocity, at a constant rate, that there is a change in joint torques that relate to the specific phase and movements occurring in that phase; with the ankle plantarflexors contributing the greatest portion of the increase in speed initially followed by the hip flexors at faster speeds (Arampatzis et al., 1999; Dorn et al., 2012). These findings are similar to the findings of Thordarson (1997) in that up to a certain velocity the ankle plantaflexors contribute to the increase in stride length which leads to increased velocity.

Further evidence of this increase in joint kinetics is seen in other research. In another study through the use of inverse dynamics, joint kinetics and muscle function were determined running at three different speeds (Belli et al., 2002). The subjects ran at 4.0 ms^{-1}, 6.0 ms^{-1}, and their maximal speed. The stance phase was the focus of this study. The results showed that as the speed progressed up to the maximal speed of the subjects that the angular velocity (rads/s) and the power (W) of the hip, knee, and ankle joint all increased with speed. Joint torques increased significantly in the hip and knee with no significant change occurring in the ankle (figure 7- different colored lines represent different velocities) (Belli et al., 2002). In Arampatzis et al (1999) increases in joint torques and powers were also seen in the knee but were seen in the ankle as well when running speed was increased. The hip joint data was not displayed. Schache et al (2011) found increases in joint torques and powers with the hip and knee having the largest increase in magnitude with speed during the terminal swing phase. Finally, in two earlier studies which also examined the changes in joint powers at all three joints collectively in relation to increases in velocity the results indicated a positive correlation between joint powers and running velocity (Fukunaga et al., 1980; Kaneko, 1990)

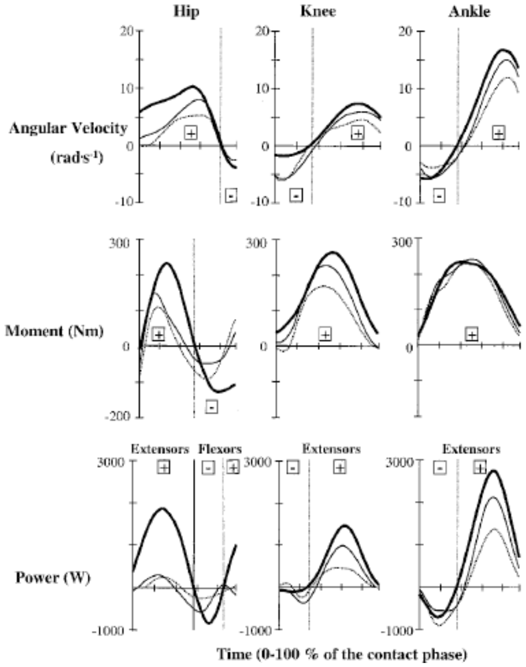

Figure 7-Time normalized, mean angular velocity, joint moment, and power curves (Belli et al., 2002)

The result for ground reaction forces also indicates a change with increasing speed. The vertical ground reaction forces increase significantly from 4.0 ms^{-1} to maximal speed (figure 8) (Belli et al., 2002; Hamill, Bates, Knutzen, & Sawhill, 1983; Munro et al., 1987). When investigating the effect that body mass and various constant speeds had on support forces in another study, it was found that at higher speeds the support ground reaction forces increased from 1.5 to 2.5 times the body weight of the subject (Figure 8) (Weyand & Davis, 2005). Dorn et al (2012) also had an increase in peak vertical ground reaction forces with an increase in speed from 2.7 BW at 3.5 ms^{-1} to 3.6 BW at 7.0 ms^{-1}. In previous studies it was also found that vertical ground reaction forces increase with an increase in running speed, but only until about 60 percent of the subject’s maximal velocity at which point they state velocity increase is a result of muscle force production or joint torque (Keller et al., 1996; Schache et al., 2011). Belli et al (2002) and

Weyand and Davis (2005) were also supported by the findings of Arampatzis et al (1999) that running velocity does influence maximum vertical velocity significantly, especially at faster velocities (<4.5 ms^{-1}) (Figures 8 and 9). These studies indicate an increase in vertical ground reaction forces with an increase in speed which are shown as well in an earlier study that performed a reexamination of running and different velocities (Hamill et al., 1983; Munro et al., 1987).

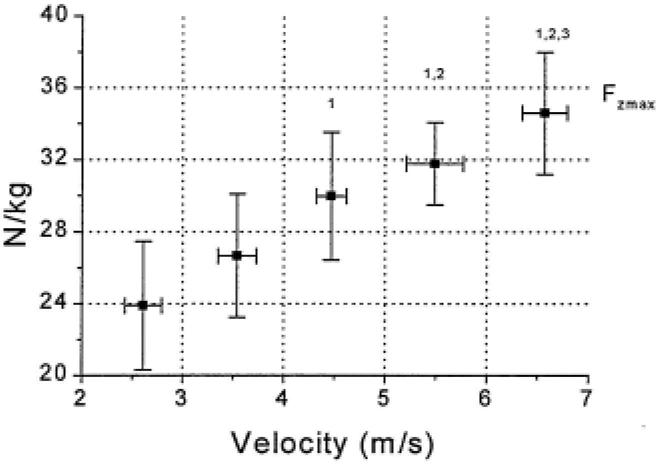

Figure 8- Peak ground forces against velocity (Weyand & Davis, 2005)

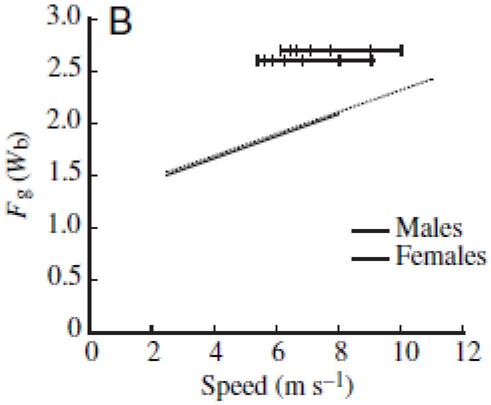

Figure 9- Maximum ground reaction forces (Arampatzis et al., 1999)

The horizontal reaction forces also increase with increased speed (figure 10). The anteroposterior forces (F_y) are also indicative of an acceleration curve with the “braking” posterior portion being smaller than the “propulsive” anterior force whereas they are equal in constant state (Belli et al., 2002; Hunter et al., 2005; Walter & Carrier, 2009). This is similar to the results found in a later study showing when increasing speed, the propulsive force is greater than the braking force (Van Caekenberghe et al., 2013).

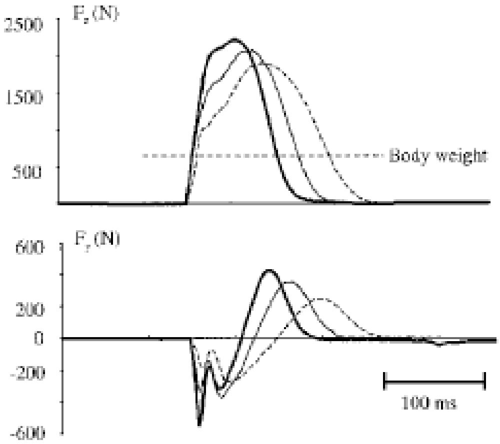

Figure 10- Mean vertical and horizontal ground reaction forces (Belli et al., 2002)

Literature in Accelerating Running

It has been stated previously that the research analyzing the acceleration phase of running is not extensive. The literature is dominated by studies investigating different constant state running speeds and the mechanical running differences between the constant states (Roberts & Scales, 2004; Van Caekenberghe et al., 2013) . Running rarely occurs at a constant state which is why the meager amount of literature on the acceleration phases is surprising. There has been growth in the study of locomotion from the walk to run transition phase but acceleration phases within running alone are not substantial. One difference between the constant-state running and accelerated running is net work must be done on the center of mass for the acceleration to occur which is in contrast to constant state locomotion where no net work is done on the center of mass. Another difference is it appears that acceleration depends on the muscles ability to rapidly shorten while also producing large propulsive forces (McGowan, Baudinette, & Biewener, 2005; Walter & Carrier, 2009). Additionally, McGowan et al (2005) stated that there is an energy shift in braking phase of stance that goes distal to proximal and vice versa in the propulsive phase. Van Caekenberghe et al (2013) performed a novel study in which they investigated the acceleration phase in both an over ground modality and treadmill modality.

The methodology and protocol for Van Caekenberghe et al's (2013) study was quite sound. The study protocol was able to measure the parameters that were desired to be tested for the study analysis. In terms of the protocol it was a relatively simple study and is similar to the methods of this study. The ten subjects ran in two modalities, over ground and on an instrumented treadmill with a force transducer. The subjects started out at a baseline of 2.0 ms^{-1} and accelerated to 7.0 ms^{-1} at different accelerations ranging between 0.0 and 3.0 ms^{-2}. Various forms of data analysis were performed using methods of inverse dynamics to arrive at the results (Visual 3-D, C-Motion, Rockville, Maryland). These protocol again are similar to the methods used in this study with the exemption of an over ground running modality (Van Caekenberghe et al., 2013).

The results stated that there were no significant changes in the sagittal plane running joint torques at the hip, knee, and ankle during the acceleration phase on an instrumented treadmill. This does not relate to the findings of a study investigating the acceleration period of turkeys which found there to be an increase in the joint torques through the acceleration periods (Roberts & Scales, 2004). Van Caekenberghe et al (2013) contribute the increase in speed to be a result of increased muscular power which does correspond with the previous literature that looked at increased constant state velocity (Arampatzis et al., 1999; Belli et al., 2002; McGowan et al., 2005; Schache et al., 2011). These results also relate to the findings of Roberts and Scales (2004). The results of the Roberts and Scales (2004) study can be seen in figures 11a-c. Van Cakeneberghe (2013) also found there to be minimal alteration of the ground reaction force orientation, contributing this to the absence of linear whole body inertia found in previous literature comparing treadmill running and over ground running (Van Caekenberghe et al., 2012). In the same study by Van Caekenberghe et al (2012), they state that braking and propulsive force amplitudes are not affected during accelerating running on a treadmill. In continued regard to ground reaction forces, the finding that average body lean is not altered during treadmill acceleration is also shown by the limited change in ground reaction forces (Van Caekenberghe et al., 2013). They state that the acceleration phase of running on the treadmill has an antero-posterior GRF curve with the anterior and posterior curves being equal which is not indicative of an acceleration phase antero-posterior GRF as found in previous studies (Hunter et al., 2005; McGowan et al., 2005; Walter & Carrier, 2009). In over ground running, the authors of this study contribute a large portion of the acceleration phase to the more anteriorly directed trunk lean which contributes to the more forward orientation of the ground reaction forces which align with previous research findings (Kugler & Janshen, 2010). Additionally, they characterize acceleration phases with smaller “braking” phases and larger “propulsive” forces in the anteroposterior ground reaction force curves which again can be seen in previous literature (Belli et al., 2002). The authors stated that joint torques in the lower extremity are not altered during acceleration; since the magnitude of the ground reaction force during the acceleration is not significantly influenced on a treadmill they arrive at this conclusion (Van Caekenberghe et al., 2013).

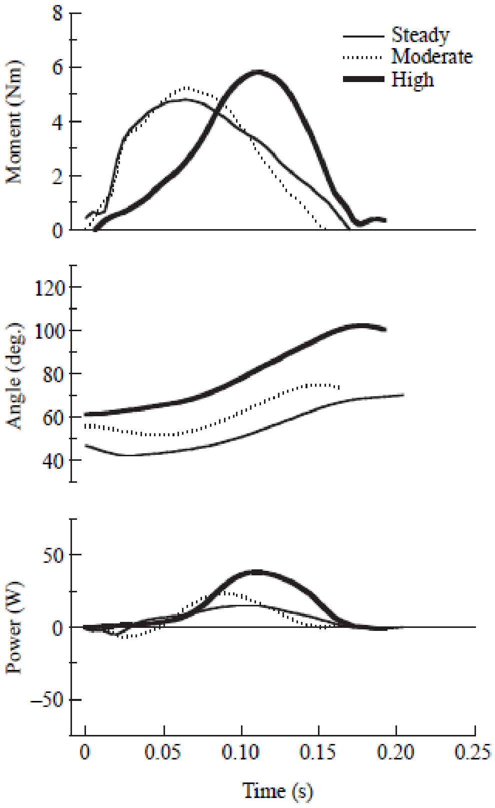

Fig.11a

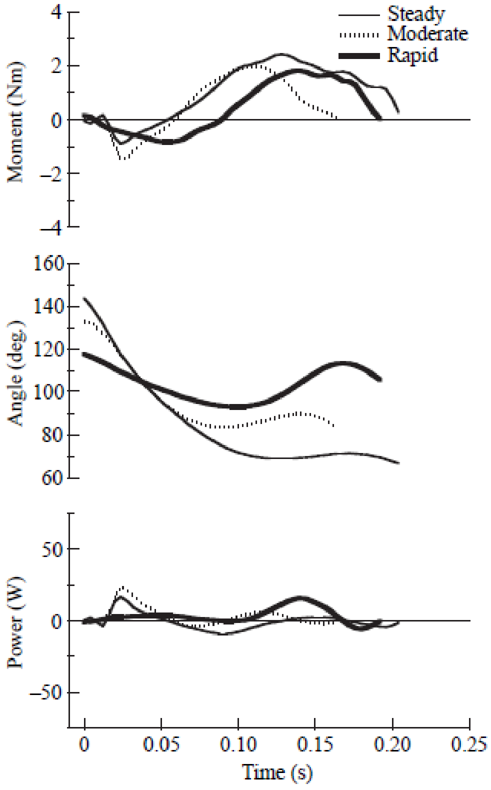

Fig. 11b

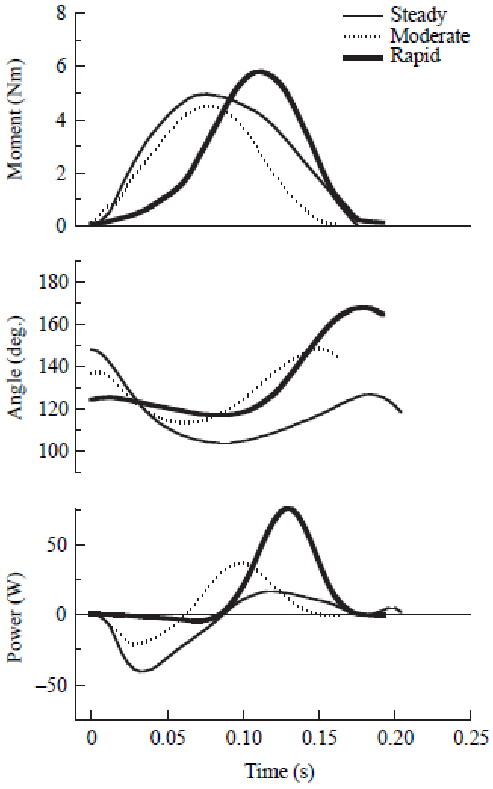

Fig.11c

Figure 11a: Hip joint torque, angle, and power for steady speed running, moderate acceleration and high acceleration. Figures 11b and 11c show the same results for the knee and ankle respectively (Roberts and Scales, 2004)

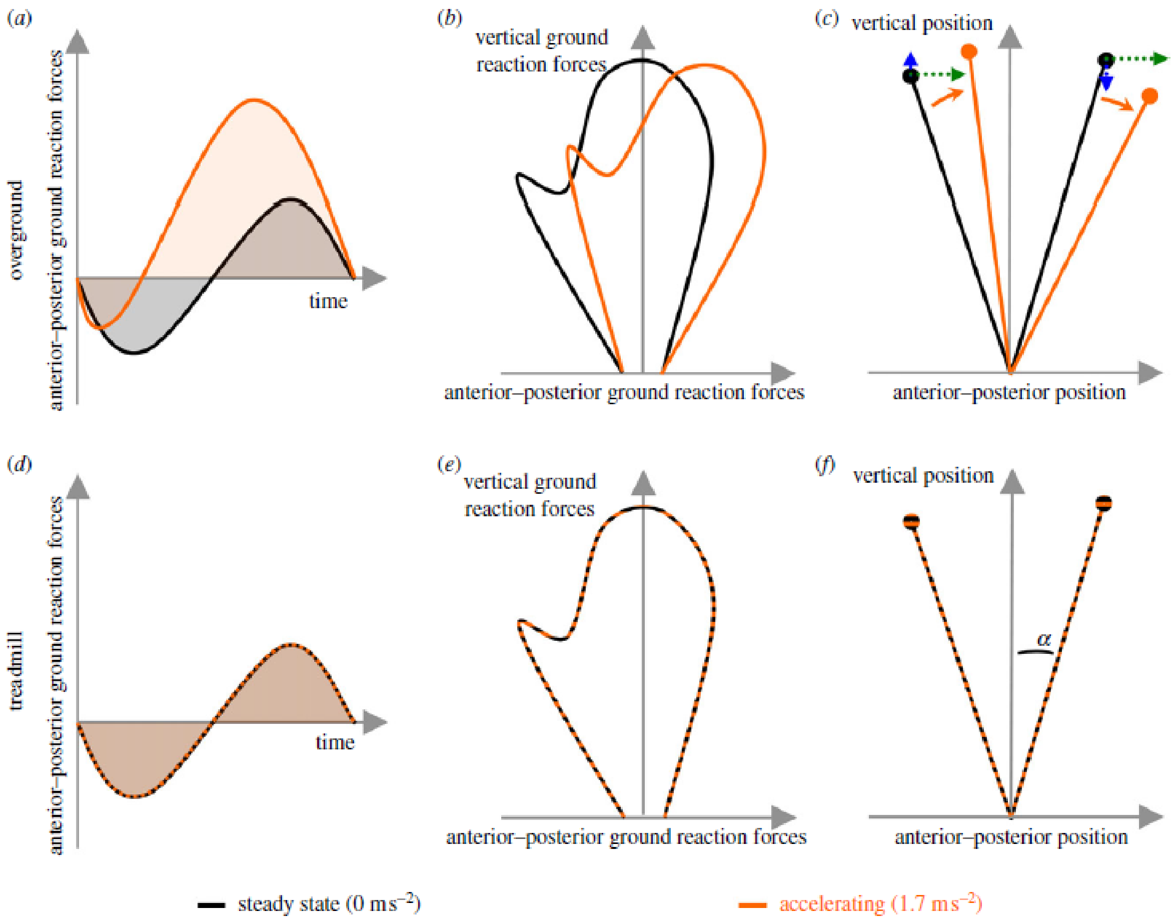

Figure 12- Ground Reaction forces during accelerated running on treadmill and overground (Van Caekenberghe et al., 2013)

Van Caekenberghe et al (2013) also compared the results of acceleration patterns on a treadmill to over ground running. In a previous study by Van Caekenberghe et al (2012) it was found that accelerated running on a treadmill is mechanically different from accelerated over-ground running. They state that there is minimal or a lack of horizontal ground reaction forces on a treadmill due to the belt moving under the body (Van Caekenberghe et al., 2012; Van Caekenberghe et al., 2013). This concept could have some validity. However, research comparing the kinematics and kinetics of treadmill running to over ground running has stated that they are similar enough to merit the use of treadmills in running biomechanics research (Riley et al., 2008). Joint torques and power curves were shown to be qualitatively similar in both over ground and treadmill running (Hamner & Delp, 2012; Riley et al., 2008). Van Caekenberghe et al (2013) again, stated that there is a minimal change of anteroposterior ground reaction forces in treadmill running (Figure 12). While the ground reaction forces were significantly reduced in treadmill running, they were present which contradicts the findings of Riley et al (2008).

Summary

The Van Caekenberghe et al study (2013) was one of the first studies in the research of running acceleration phases. The methods of the study were sound in terms of testing the parameters desired but it does not line up with the findings of previous research in velocity. The aforementioned studies that researched increased constant state velocity indicated there was a significant change in the joint torques and joint powers when the running speed was increased. Granted, these studies were researching increases in constant state velocity, but since accelerated running is a change in velocity over time if one positively accelerates their velocity is increasing so the mechanics found should be similar in the acceleration phase of running. These studies have provided very pertinent information to the biomechanics involved in running but running is rarely held at a constant velocity for an entire period of time. With that said, the focus of this study will be the investigation of the acceleration phase in running in order to add to the scientific literature on acceleration phases. We will perform this study on an instrumented treadmill with force transducers and though there is much debate on the ability to generalize the data found in treadmill studies to over ground running, based on the findings of Riley et al (2008) we find this modality to be validated.

Based on the previous research investigating running biomechanics, including velocity related changes in running biomechanics, it was hypothesized that lower extremity, sagittal plane joint torques and joint powers would positively and linearly increase throughout the acceleration phase of running..

CHAPTER 3: METHODOLOGY

Based on the previous research investigating running biomechanics, including velocity related changes in running biomechanics, it was hypothesized that lower extremity, sagittal plane joint torques and joint powers would positively and linearly increase throughout the acceleration phase of running. The purpose of this study was to quantify lower extremity joint torques and powers during constant speed running and during running while accelerating at two rates of acceleration between a baseline velocity of 2.50 ms^{-1} to 6.00 ms^{-1}. The rates of acceleration were 0.40 ms^{-2} and 0.80 ms^{-2}. The general procedures for the study will be outlined in the following sections. The testing consisted of four trials in one session at the Human Movement Analysis Lab at East Carolina University. The subjects performed trials at two different rates of acceleration two times each in case there is an error in the first trial. Kinematic and kinetic data were collected and analyzed. From these data the differences in joint torques and powers at each rate of acceleration were determined.

Participants

The participants were recruited from the East Carolina University student body and the citizens of the city of Greenville through the use of fliers and classroom announcements. They were then screened to see if they meet all inclusion criteria. 15 participants (n = 8 females), all of whom were experienced runners (running at least 6 miles per week), were selected to perform the protocol for the study. Experienced runners were desired in hopes they would be more comfortable and capable with running at higher speeds on an instrumented treadmill. The participants were young and between the ages of 18-22. Additional inclusion criteria were that the participants had a body mass index of less than 28.0 kgm^{-2}, as well as free of pain when they run and in daily activities. The participants must not have had any history of severe lower extremity injuries so as to not have any effect on their normal running gait. The protocol was approved by the East Carolina University Institutional Review Board and all subjects signed an informed consent.

Inclusion Criteria

The following criteria were met to participate in this study:

1. Young experienced runners running at least 10 miles per week between the ages of 18-25.
2. No recent lower extremity injuries or musculoskeletal disorders that affected running performance presently.
3. Body Mass Index of less than 28.0 kgm^{-2}.
4. Signed written informed consent.
5. Free of pain when running and in daily activities.
6. Must be ostensibly healthy having no history of cardiovascular disorders or problems.

Exclusion Criteria

The following criteria would result in the inability to participate in the study:

1. A recent history of musculoskeletal injuries or disorders in the lower extremity.
2. Presently ill or suffering from cardiovascular, pulmonary, or neurological diseases or other major diseases that would affect the ability to run.
3. Smoking of cigarettes, cigars, or any form of tobacco.

Instruments

The accelerated running trials took place on a Bertec instrumented split belt treadmill (Bertec Corporation, Columbus, Ohio) with force transducers located under the deck. Running kinematic data was captured with Qualisys Oqus 300 cameras (Qualisys Medical AB, Gothenberg, Sweden). Qualisys Track Manager (Innovision Systems, Columbiaville, Maryland) software was used to collect the kinematic and kinetic data at an analog frequency of 2400 Hz and a motion capture frequency of 240 Hz. The kinematic and kinetic data was then analyzed using Visual 3D Software (C-Motion Inc., Rockville, Maryland) and Microsoft Excel (Microsoft Corporation, Redmond, Washington).

Protocol

The testing took place in the Human Movement Analysis Laboratory in the Department of Physical Therapy at East Carolina University (Greenville, North Carolina). The participants were measured for their height and weight and also asked their age. The data collection occurred in one session and lasted approximately one to one and a half hours. Participants changed into tight fitting compression shorts to prevent movement of the markers that were placed on both legs. They wore their own running shoes to ensure running comfort when performing the protocol.

Before performing the acceleration trials, reflective markers were placed on the lower extremity using the locations of the Modified Helen Hayes marker set on both legs of the participants. The location of the markers were: right and left posterior superior iliac spine (PSIS), right and left iliac crest, right and left anterior superior iliac spine (ASIS), right and left greater trochanter, medial and lateral knee, medial and lateral malleoli, heel, first and fifth metatarsal heads, a thigh plate consisting of four markers was placed on the lateral side of the thigh, a shank plate consisting of four markers was placed on the lateral side of the leg, and finally a foot plate with three markers (two on the lateral side, one on the medial) was placed on the superior side of the foot. A five second calibration was taken. The subject stood completely still with their arms folded across their chest so no markers were blocked by the arms. After the calibration trial, the iliac crests, greater trochanters, knee, malleoli, and metatarsal head markers were removed.

The participants performed a series of four trials on the instrumented treadmill- unless an error occurred in which additional trials were collected. The participants had adequate time to become familiar with the treadmill so they could achieve their routine running cadence and gait at the designated velocity. They had approximately 5 minutes to warm up on the treadmill prior to the testing protocol began and could have requested more time if it was needed. After the warmup, they had a practice trial of the 0.40 ms^{-2} acceleration trial. Another practice trial was given after they performed the 0.40 ms^{-2} trials for 0.80 ms^{-2}. The participants then performed the four trials which lasted approximately 45-60 seconds each. They began running at a velocity of 2.5 ms^{-1} for approximately 20 seconds at which point they then accelerated to a maximum velocity of 6.0 ms^{-1}. The partcipants accelerated at two different rates for two trials each for a total of four trials. The rates of acceleration were, 0.40 ms^{-2} (A1) and 0.80 ms^{-2} (A2). They would then accelerate at the designated rates to a velocity of 6.0 ms^{-1} at which point they ran for approximately 10-15 seconds before being brought back to the initial velocity of 2.5 ms^{-1}. Sufficient rest was given between trials so fatigue was not a confounding variable. The initial and maximal velocities along with the rates of acceleration were determined through pilot testing as to what was reasonable for the participants to perform on the instrumented treadmill. It was also desired to have two rates of acceleration that were sufficiently different from one another. The rates of acceleration were within the range that Van Caekenberghe et al (2013) used.

Data Reduction

The data from the trials were processed in Qualisys Track Manager software. The reflective markers placed on the participants were then identified leading to the creation of a three dimensional model using a rigid segment system as the basis for model creation. The three dimensional model was created in Visual 3D software using the static calibration model. This model was then applied to the motion trials and kinetic data was produced. Visual 3D was able to find the location of joint centers which were calculated using 50 percent of the distance between the reflective markers on the medial and lateral markers of the knee and ankle and 25 percent from the greater trochanters for the hip, segment centers of mass were based on anthropometric data, and definition of the local coordinate system. Once all kinematic data were determined, the data was processed with a low pass digital filter to remove high frequency position error. A standard cut-off frequency of 6 Hz was used in the second order-low pass Butterworth digital filter and 45 Hz cutoff was used for the GRF data. Linear velocities and accelerations of the joint centers and segmental mass centers were then calculated. Ankle, knee, and hip joint angular velocities were calculated from the processed positional data.

Inverse dynamics methods using linear and angular Newtonian equations of motion were utilized through the use of rigid segment models of the lower extremity (thigh, leg, and foot) to calculate the joint reaction forces and torques from the measured ground reaction forces (N), center of pressure, standardized segment anthropometrics, and the kinematic position and acceleration data. The Newtonian mechanical analysis to calculate the joint reaction forces and torques is displayed in Figure 13 as a series of free body diagrams of the foot, leg, and thigh.

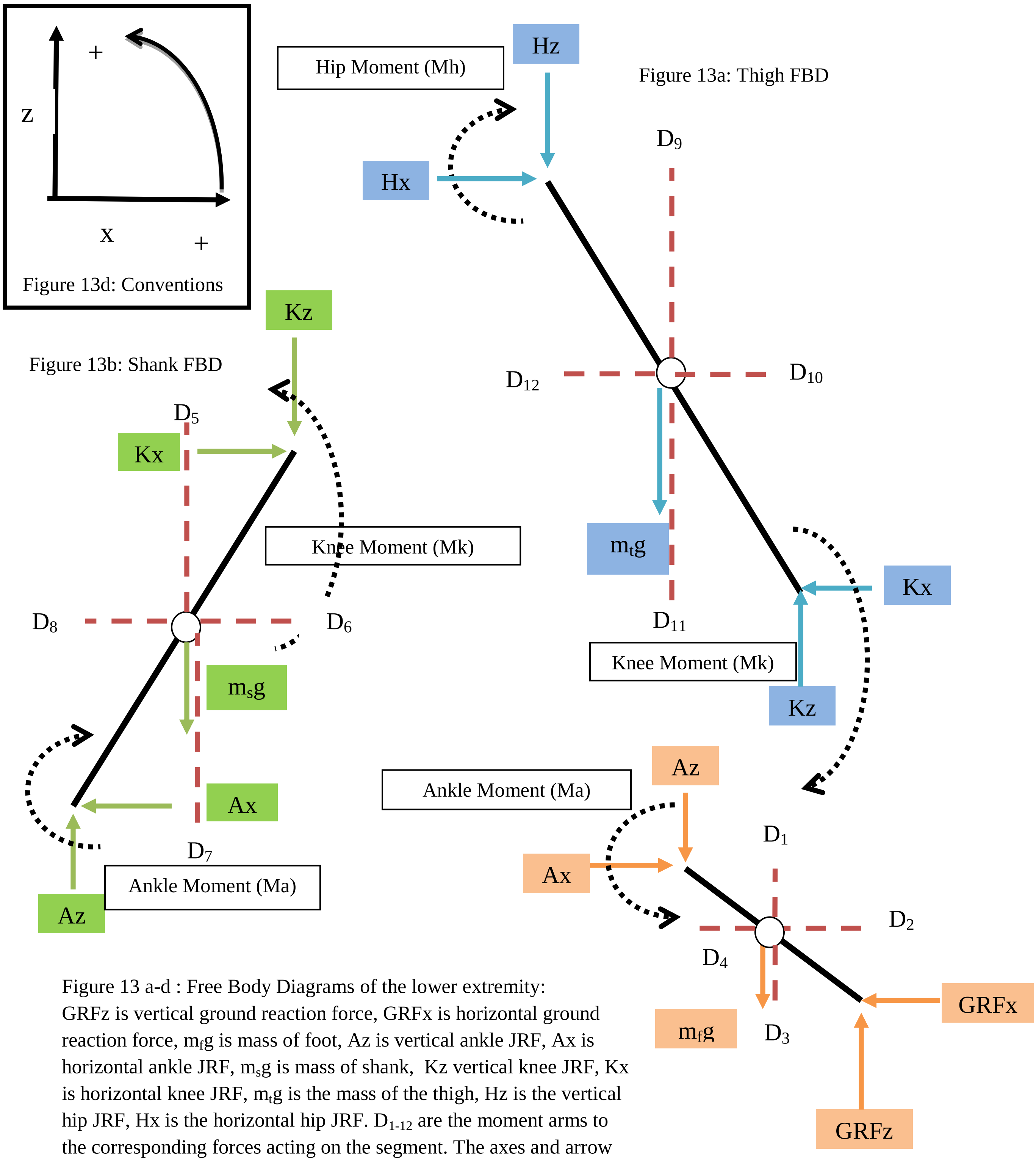

Figure 13d: Conventions

Figure 13a: Thigh FBD

Figure 13b: Shank FBD

Figure 13c: Foot FBD

Figure 13 a-d : Free Body Diagrams of the lower extremity: GRFz is vertical ground reaction force, GRFx is horizontal ground reaction force, $m_f g$ is mass of foot, Az is vertical ankle JRF, Ax is horizontal ankle JRF, $m_s g$ is mass of shank, Kz vertical knee JRF, Kx is horizontal knee JRF, $m_t g$ is the mass of the thigh, Hz is the vertical hip JRF, Hx is the horizontal hip JRF. $D_{1\text{-}12}$ are the moment arms to the corresponding forces acting on the segment. The axes and arrow in the top left corner indicate the linear and angular conventions are positive. Figure 13d is the conventions diagram to indicate which directions are positive.

The variables seen in the free body diagrams are the values used by the Visual 3D software to calculate the joint reaction forces (JRF) and then joint torques using the following equations. The calculations started at the foot since that is the segment in contact with the force plate which provides the only measured external force which then translates proximally up to the knee and then the hip. The basis for these calculations is Newton's Second Law:

$$F = ma$$

Where F is the force, m is the mass of the segment, and is the acceleration of the object and the forces in the various planes will be summed. The determination of the variables being positive or negative is based on the coordinate system found in the FBD (Figure 10) and the conventions for torque will be counterclockwise is positive and clockwise is negative.

The equation for the vertical joint reaction force at the ankle (Az) is represented as such:

$$GRFz - m_f g + Az = m_f a_{zf}$$

Where GRFz represents the ground reaction forces in the vertical direction, $m_f g$ represents the product of the mass of the foot and the acceleration due to gravity, Az is the ankle vertical joint reaction force, m_f is the mass of the foot as calculated from the anthropometric proportions, a_{zf} is the vertical acceleration of the foot.

The equation for the horizontal joint reaction force at the ankle is represented as such:

$$-GRFx + Ax = m_f a_{xf}$$

Where GRFx represents the ground reaction forces in the horizontal direction, Ax represents represents the ankle joint reaction force in the horizontal direction, and a_{xf} represents the horizontal acceleration of the foot.

The previous two equations will calculate the variables necessary to determine the torque of the ankle with the use of the equation:

$$T = I\alpha$$

Where T is the calculated torque, I represents the moment of inertia, and α represents the angular acceleration of the foot. The sum of the products of the four forces acting on the foot and their respective moment arms will then be used to determine the torque of the ankle (Ma).

The equation to calculate the torque of the ankle (Ma) is represented as such:

$$AxD_1 + GRFzD_2 - GRFxD_3 + AzD_4 + Ma = I_f\alpha_f$$

Where D_1-D_4 represent the moment arms of the ground reaction forces in both the vertical and horizontal directions (D_2 and D_3 respectively) and the ankle joint reaction forces in the horizontal and vertical direction (D_1 and D_4 respectively) from the center of mass of the foot, and Ma represents the ankle torque. I_f represents the moment of inertia of the foot, and α_f represents the angular velocity of the ankle in the sagittal plane.

The knee and hip will follow similar calculations to find the joint reaction forces in the horizontal and vertical direction first, followed by the calculation for the knee torque and hip torque.

The equation for the vertical joint reaction of the knee is represented as such:

$$Az - m_s g + Kz = m_s a_{zs}$$

Where Az represents the vertical ankle joint reaction force, $m_s g$ represents the product of the mass of the shank and the acceleration due to gravity, Kz is the vertical knee joint reaction, and m_s is the mass of the shank, and a_{zs} is the vertical acceleration of the shank.

The equation for the horizontal joint reaction force of the knee is represented as such:

$$Ax + Kx = m_s a_x$$

Where Ax represents the horizontal ankle joint reaction force, Kx represents the horizontal knee joint reaction force, m_s represent the mass of the shanks, and a_x represents the horizontal acceleration of the shank.

With the calculated JRFs at the knee the torque will be calculated and represented as such:

$$KxD_5 - KzD_6 - AxD_7 - AzD_8 + Mk = I_s\alpha_s$$

Where $D_5 - D_8$ represent the moment arms of the knee joint reaction forces in the horizontal and vertical direction (D_5 and D_6 respectively) and the ankle joint reaction forces in the horizontal and vertical directions (D_7 and D_8 respectively) from the center of mass of the shank. Mk represents the knee torque, I_s represents the moment of inertia of the shank, and α_s represents the angular acceleration of the shanks in the sagittal plane.

The equation for the vertical hip joint reaction force is represented as such:

$$Kz - m_t g + Hz = m_t a_{zt}$$

Where Kz represents the vertical knee joint reaction force, $m_t g$ is the product of the mass of the thigh and the acceleration due to gravity, Hz is the vertical hip joint reaction force, m_t is the mass of the thigh, a_{zt} is the vertical acceleration of the thigh.

The equation for the horizontal hip joint reaction force is represented as such:

$$Kx + Hx = m_t a_{xt}$$

Where Kx represents the horizontal knee joint reaction force, Hx is the horizontal hip joint reaction force, m_t is the mass of the thigh, and a_{xt} represents the horizontal acceleration of the thigh.

The equation for the hip torque is represented as such:

$$HxD_9 + HzD_{12} – KxD_{11} + KzD_{10} + Mh = I_t \alpha_t$$

Where $D_9 – D_{12}$ represent the moment arms of the knee joint reaction forces in the horizontal and vertical direction (D_{10} and D_{11} respectively) and the hip joint reaction forces in the horizontal and vertical directions (D_9 and D_{12} respectively) from the center of mass of the shank. Mh represents the hip torque, I_t is the moment of inertia of the thigh, α_t is the angular acceleration of the thigh.

The aforementioned equations and methods are the basis by which Visual 3D software calculated the the joint torques. Joint powers will be calculated from the product of the calculated joint torques and calculated joint angular velocities (Elftman, 1940; Johnson & Buckley, 2001; Winter, 1983). Peak hip, knee, and ankle, sagittal plane joint torques and powers will then be derived at the hip, knee, and ankle joints. The peak value for each step- as determined through a Visual 3D pipeline command- of each participant was entered in to a Microsoft Excel (Microsoft Corporation, Redmond, Washington) spreadsheet for each step taken during the acceleration period; as well as ten steps before acceleration and ten steps after acceleration. The acceleration period was determined through the use of a heel switch which was activated at the start of the acceleration of the treadmill belt. It remained activated until the acceleration period ended. The peak values for the participants of both trials in each condition were then used to find the mean peak values for hip, knee, and ankle joint torques as well as hip, knee, and ankle joint powers. This allowed for an overall representation of all participants but also a view of each individual subject and the inter-subject variation that occurred from the differences in how they ran during the trials.

Statistical Analysis

A set of correlation coefficients and statistical regressions were performed in which the peak hip, knee, and ankle joint torques and powers at each step were correlated and regressed to the step number during the acceleration phase to identify the relationship between these variables and step number. A 95 % confidence interval was used to determine if there were significant differences between the different joints as well as the two conditions. To account for the uneven number of steps, we found the mean peak value of each step for all participants and used those for the regression analysis. The significance level was set at $p < 0.05$.

CHAPTER 4: RESULTS

Based on the previous research investigating running biomechanics, including velocity related changes in running biomechanics, it was hypothesized that lower extremity, sagittal plane joint torques and joint powers would positively and linearly increase throughout the acceleration phase of running. The purpose of this study was to quantify lower extremity joint torques and powers during constant speed running and during running while accelerating at two rates of acceleration (0.40 ms^{-2} and 0.80 ms^{-2}) between a baseline velocity of 2.50 ms^{-1} to 6.00 ms^{-1}. The results section will be divided in to the following sections, demographics, hip joint torques, knee joint torques, ankle joint torques, hip joint powers, knee joint powers, and ankle joint powers, regression analysis, with both conditions A1 and A2 presented in each section and then a summary.

VIDEO LINKS:

Participant Running Protocol

Visual 3D Model And Figures

I). *Demographics*

The participants were recruited from the East Carolina University student body and were between the ages of 18 - 22 (mean age of 19.7 years ± 1.3 years). The sample consisted of 15 (n = 8 females) healthy, young runners with an average BMI of 22.0 kgm^{-2}.

II). *Hip Joint Torques*

Figure 14 shows an individual, representative, curve of the hip joint torque with values highlighting the beginning, middle, and end of the acceleration period respectively for one participant. This figure shows the progression of the increase in the magnitude of the hip torque through the acceleration period and is representative of both conditions. The phase correlations for step number and torque for A1 and A2 are presented in Table 1. The pre-acceleration, acceleration, and post acceleration correlations are 0.128, 0.993 and 0.118 for A1 respectively, with the acceleration period being significant ($p < 0.05$). For A2, the correlations were 0.097, 0.941, and 0.507 for the pre-acceleration, acceleration, and post-accelerations respectively, and the acceleration and post-acceleration periods were significant ($p < 0.05$). This shows that there is a strong, direct relationship between step number and torque through the acceleration period and a moderate relationship in the post-acceleration period.

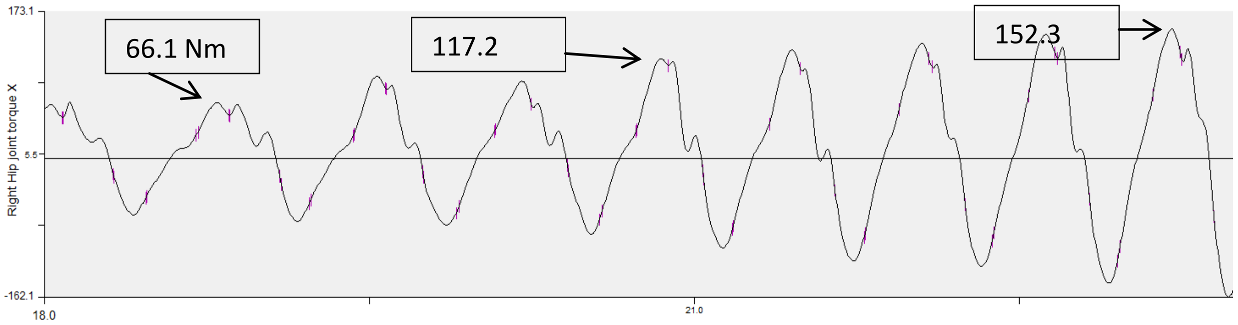

Figure 14: Representation of right leg hip joint torque during acceleration phase (S3.C1.T1)

Condition	Pre-Acceleration	Acceleration	Post-Acceleration
A1	0.128	**0.993***	0.118
A2	0.097	**0.941***	**0.507***

Table 1: A1 and A2 Hip extensor torque correlation coefficients between maximum stance phase torque and step number during the acceleration phase; * $p<0.05$

Figure 15 shows the mean peak values of all participants of the pre-acceleration, acceleration, and post-acceleration phases for A1. A linear regression beta weight of best fit was calculated for the acceleration period, $y = 3.2268x$ with an R^2= of 0.987 ($p < 0.05$) showing an increase in the magnitudes of the three joint torques during the acceleration phase. The pre-acceleration and post-acceleration regression beta weights were calculated to be $y = 0.02x$, $R^2 = 0.0163$ and $y = 0.0407x$, $R^2 = 0.014$, respectively, indicative of no change in the torque values. Figure 16 shows the mean peak values of all participants of the three phases for A2 and had a similar trend to that of A1. The linear regression beta weight of best fit was calculated to be $y = 3.801x$ with an $R^2 = 0.8863$ ($p < 0.05$) which is indicative of a significant increase through the acceleration period. The pre-acceleration and post-acceleration regression beta weights were calculated to be $y = 0.015x$, $R^2 = 0.0094$, and $y = 0.2372x$, $R^2 = 0.2566$, respectively, again indicating no change in hip joint torque magnitudes in the constant states.

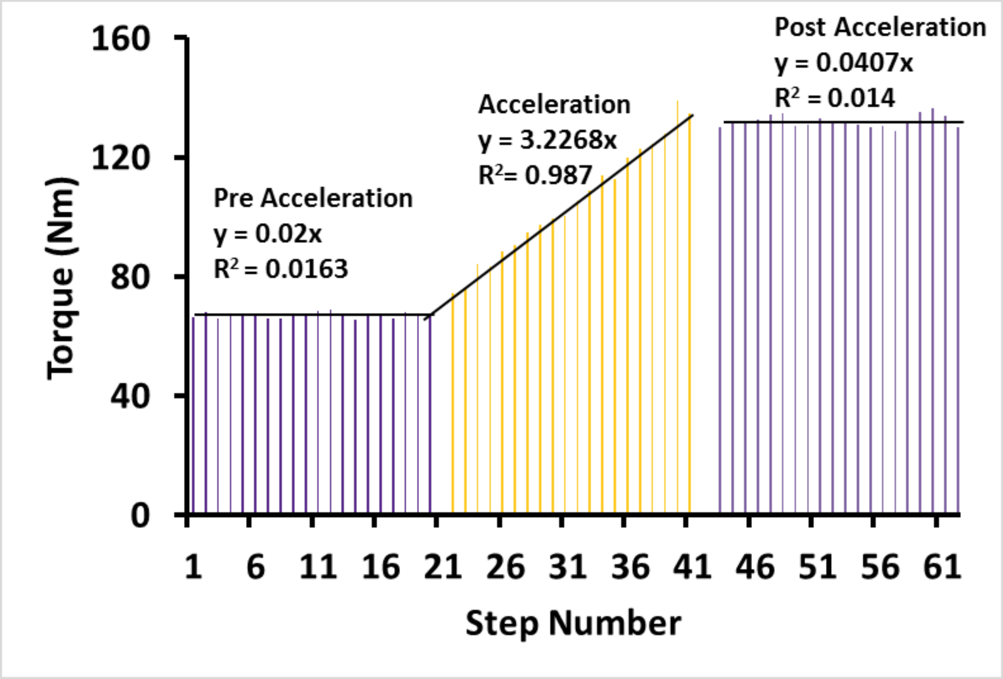

Figure 15: A1 Peak mean hip extensor torque during pre-, post- acceleration phases

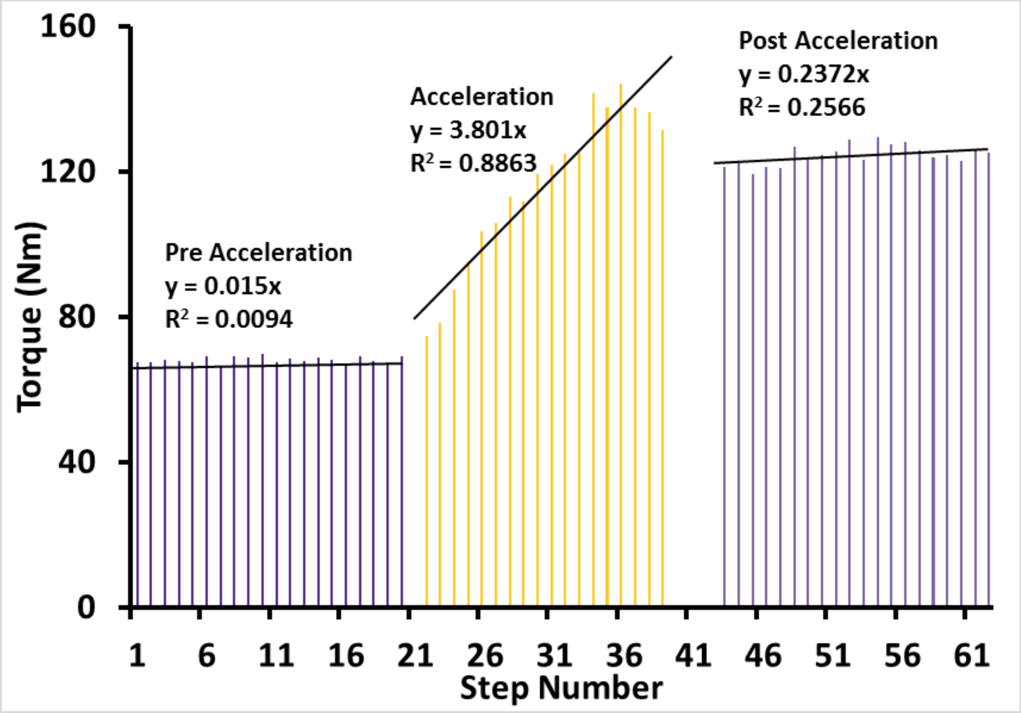

Figure 16: A2 Peak mean hipextensor torque during pre-, post- acceleration phases

III). *Hip Joint Powers*

Figure 17 shows an individual, representative curve of the hip power with values highlighting the beginning, middle, and end of the acceleration period respectively for one participant. This figure shows the progression of the increase in the magnitude of the concentric hip power through the acceleration period and is representative of both conditions. The phase correlations for A1 and A2 are presented in Table 2. The step number and torque correlations for the pre-acceleration, acceleration, and post-acceleration periods are 0.518, 0.989, and 0.482 respectively for A1 with the pre-acceleration and acceleration periods being significant ($p < 0.05$). For A2 the step correlations were calculated to be 0.132, 0.917, and 0.104 for the pre-acceleration, acceleration, and post-acceleration period respectively, with the acceleration period being significant ($p < 0.05$). This shows that there is a strong, direct relationship between step number and torque through the acceleration period for hip joint powers and a moderate, positive relationship in the pre-acceleration period for A1 and A2.

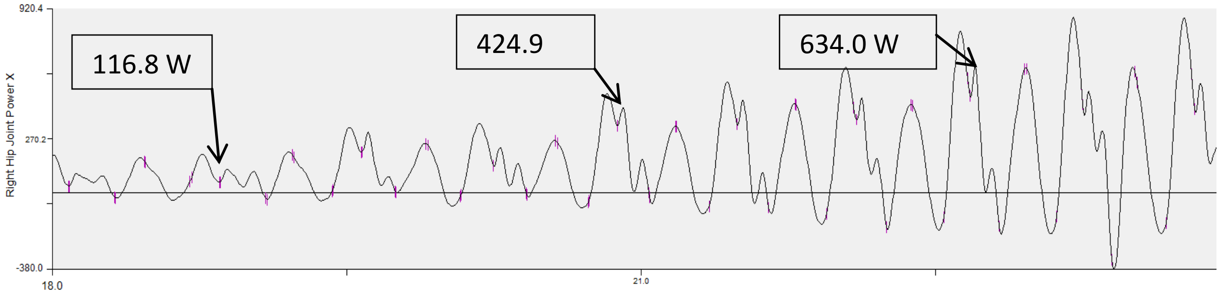

Figure 17: Representation of right leg hip joint power during acceleration phase (S3.C1.T1)

Condition	Pre-Acceleration	Acceleration	Post-Acceleration
A1	**0.518***	**0.989***	0.482
A2	0.132	**0.917***	0.104

Table 2: A1 and A2 Hip power correlation coefficients between maximum concentric stance phase power and step number during the acceleration phase; * $p<0.05$

Figure 18 shows the mean peak hip power values of all participants of the three phases for A1. A linear regression beta weight of best fit was calculated for the acceleration period to be, $y = 12.846x$, with an R^2= of 0.9786 ($p < 0.05$). This indicates that there is a significant increase in concentric extensor power magnitude at the hip joint when accelerating. The pre-acceleration and post-acceleration regression beta weights for A1 were calculated to be $y = 0.8654x$, $R^2 = 0.2682$ and $y = 1.7265x$, $R^2=0.2323$ respectively, indicating no change in joint power magnitude during the constant state periods. Figure 19 shows the mean peak values for hip joint power in A2. A linear regression beta weight of best fit was calculated for the acceleration period, $y = 16.447x$ and $R^2 = 0.8411$ ($p < 0.05$). The pre-acceleration and post-acceleration regression beta weights for A2 were calculated to be $y = -0.3746x$, $R^2 = 0.0174$ and $y = -0.7173x$, $R^2 = 0.0108$ respectively, indicative of no change in hip joint powers during the constant state periods.

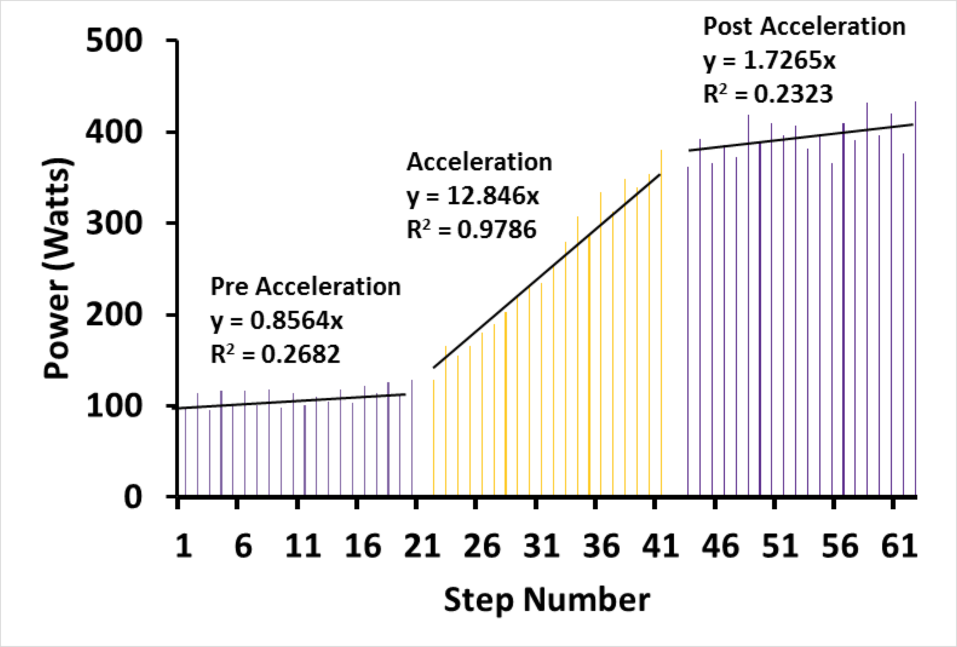

Figure 18: A1 Peak mean sagittal plane concentric hip power during pre-, post- acceleration phases

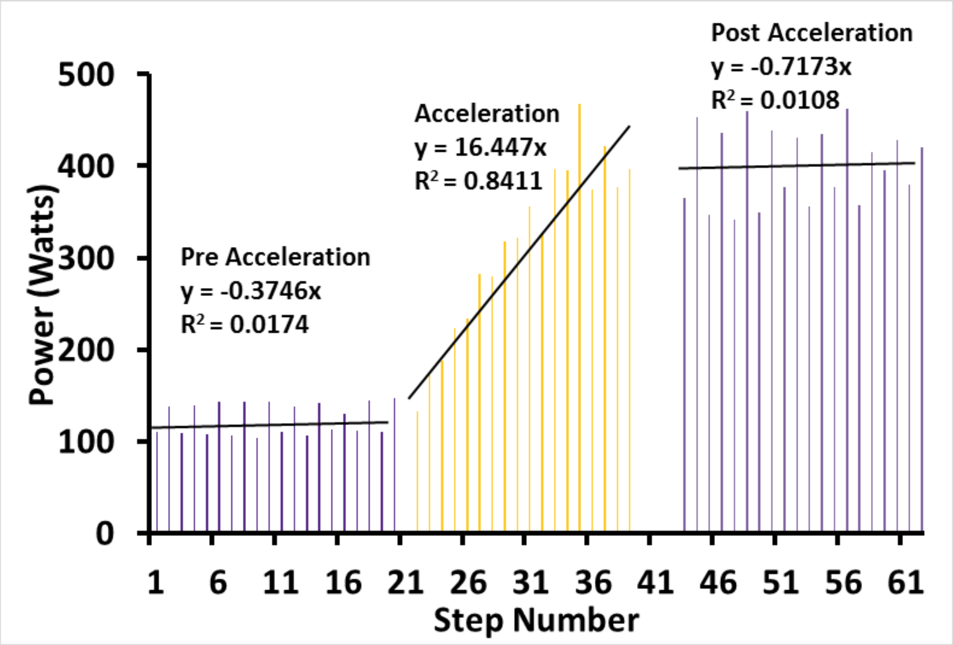

Figure 19: A2 Peak mean sagittal plane, concentric hip power during pre-, post- acceleration phases

IV). *Knee Joint Torques*

Figure 20 shows an individual, representative curve of the knee torque with values highlighting the beginning, middle, and end of the acceleration period respectively for one participant and is representative of both conditions. This figure shows the increase in the magnitude of the knee torque from the beginning to the end of the acceleration period. The correlation between step number and torque for the three phases in A1 and A2 are presented in Table 3. The pre-acceleration, acceleration, and post-acceleration periods were -0.316, 0.896, and -0.213 respectively for A1, with the acceleration period being significant ($p < 0.05$). For A2, the step correlations were calculated to be 0.063, 0.946, and -0.014 for the pre-acceleration, acceleration, and post-acceleration period respectively, with the acceleration period being significant ($p < 0.05$). This shows that there is a strong, direct relationship between step number and torque through the acceleration period at the knee joint.

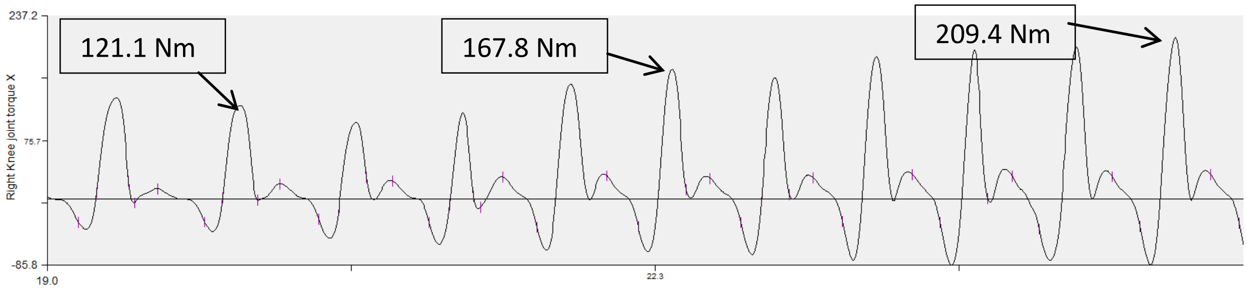

Figure 20: Representation of right leg knee joint torque during acceleration phase (S07.C2.T1)

Condition	**Pre Accel**	**Accel**	**Post Accel**
A1	-0.316	**0.896***	-0.213
A2	0.063	**0.946***	-0.014

Table 3: A1 and A2 Knee extensor torque correlation coefficients between maximum stance phase torque and step number during the acceleration phase; * $p<0.05$

Figure 21 shows the mean peak values of all participants for the three phases for A1. A linear regression beta weight of best fit was calculated for the acceleration period, $y = 0.8089x$, with $R^2 = 0.8021$ ($p < 0.05$) indicating a significant increase during the acceleration period. The pre-acceleration and post-acceleration regression beta weights were calculated to be $y = -0.0678x$, $R^2 = 0.010$ and $y = -0.0989x$, $R^2 = 0.0453$ respectively, indicating no change in the magnitude of knee joint torque through the constant state periods. Figure 22 shows the mean peak values for knee joint torque in A2. A linear regression beta weight of best fit was calculated for the acceleration period, $y = 1.162x$ and $R^2 = 0.8948$ ($p < 0.05$) which is also indicative of a significant increase in the acceleration period. The pre-acceleration and post-acceleration regression beta weights were calculated to be $y = 0.0232x$, $R^2 = 0.0002$ respectively, and indicate no change in joint torque magnitude during the constant state periods.

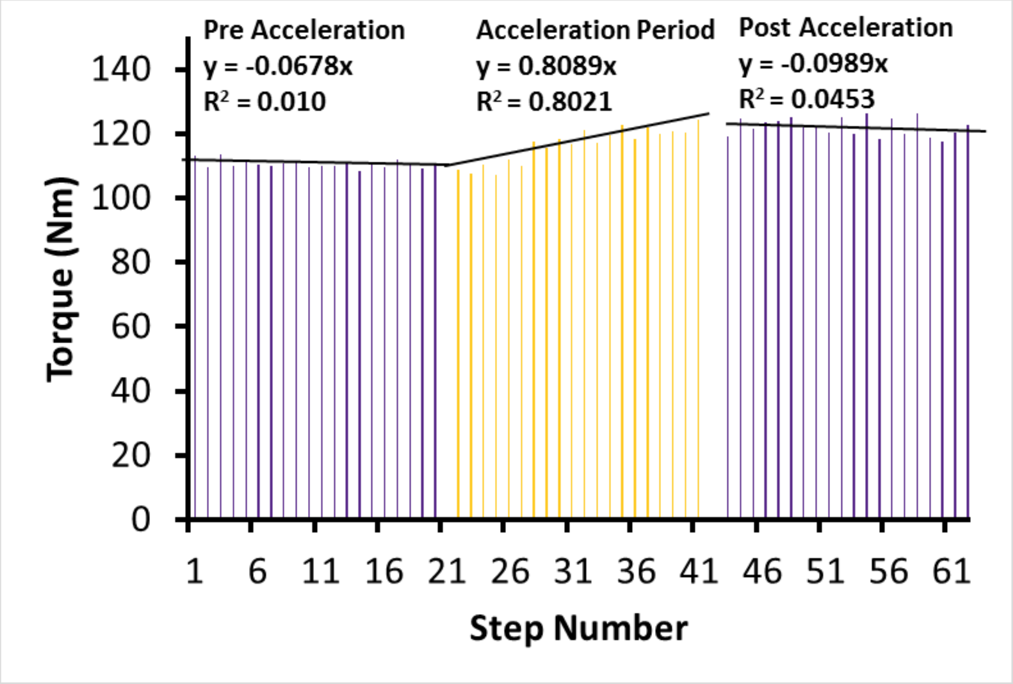

Figure 21: A1 Peak mean knee extensor torque during pre-, post- acceleration phases

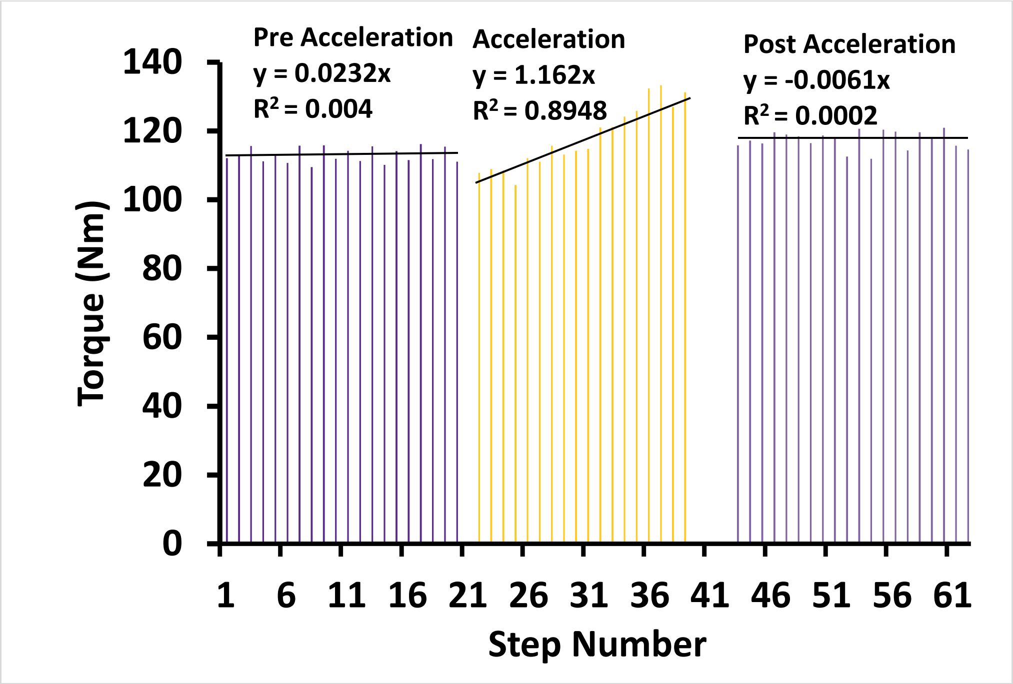

Figure 22: A2 Peak mean knee extensor torque during pre-, post- acceleration phases

V). *Knee Joint Powers*

Figure 23 shows an individual, representative curve of the knee power with values highlighting the beginning, middle, and end of the acceleration period respectively for one participant. This figure shows the increase in the magnitude of the concentric knee power through the acceleration period and is representative of both conditions. The correlation between step number and concentric knee joint power for the three phases in A1 and A2 are presented in Table 4. For the pre-acceleration, acceleration, and post-acceleration they are -0.027, 0.966, and 0.196 respectively, with the acceleration period being significant ($p < 0.05$). For A2, the step correlations were calculated to be -0.126, 0.897, and 0.250 for the pre-acceleration, acceleration, and post-acceleration period respectively, with the acceleration period being significant ($p < 0.05$). This shows that there is a strong, direct relationship between step number and concentric power through the acceleration period for the knee joint.

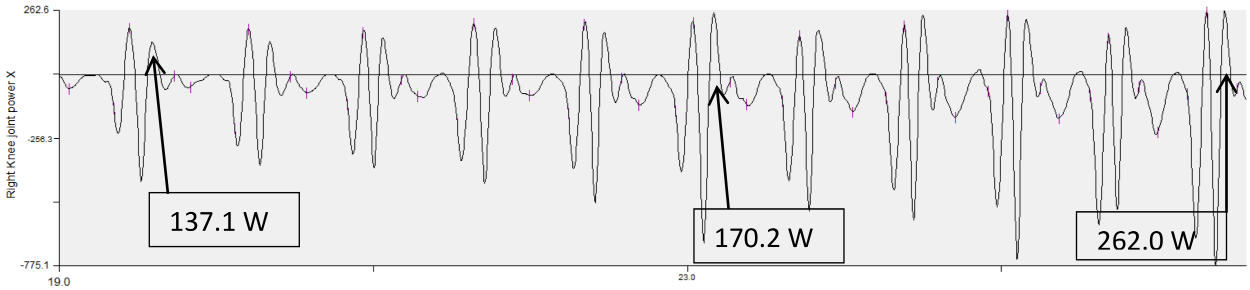

Figure 23: Representation of right leg knee joint power during acceleration phase (S06.C1.T1)

Condition	Pre Accel	Accel	Post Accel
A1	-0.027	**0.966***	0.196
A2	-0.126	**0.897***	0.250

Table 4: A1 and A2 Knee concentric power correlation coefficients between maximum stance phase power and step number during the acceleration phase; * $p<0.05$

Figure 24 shows the mean peak values of all participants for the three phases for A1. A linear regression beta weight of best fit was calculated for the acceleration period to be $y = 6.0547x$, with an R^2= of 0.9328 ($p < 0.05$) indicating a significant increase power magnitude during the acceleration period. The pre-acceleration and post-acceleration regression beta weights were calculated to be $y = -0.0324x$, $R^2 = 0.0007$ and $y = 0.4092x$, $R^2 = 0.0382$, indicating no change in the constant state periods. Figure 25 shows the mean peak values for knee joint power in A2. A linear regression beta weight of best fit was calculated for the acceleration period, $y = 9.1621x$, and $R^2 = 0.804$ ($p < 0.05$), indicating a significant increase in the acceleration period. The pre-acceleration and post-acceleration regression beta weights were calculated to be $y = -0.1386x$, $R^2 = 0.0158$, and $y = 0.501x$, $R^2 = 0.0623$ respectively, again indicating no change in the knee joint powers during the constant state periods.

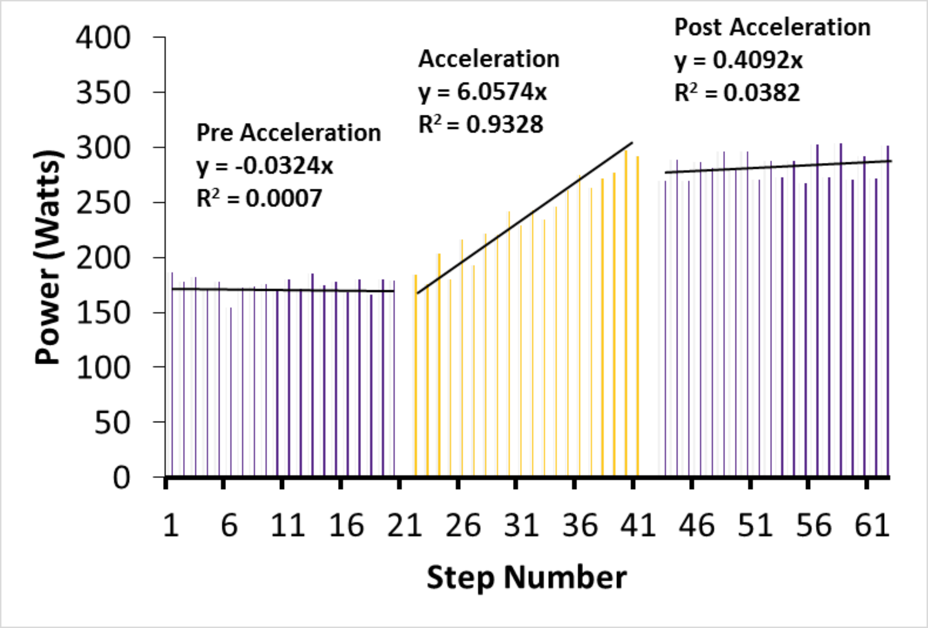

Figure 24: A1 Peak sagittal plane, concentric knee joint power during pre-, post- acceleration phases

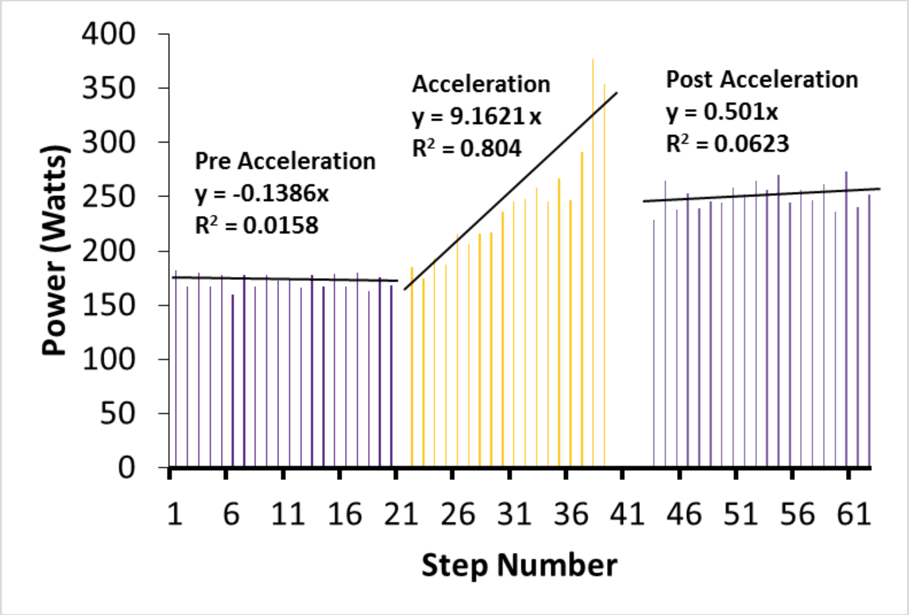

Figure 25: A2 Peak sagittal plane, concentric knee joint power during pre-, post- acceleration phases

VI). *Ankle Joint Torques*

Figure 26 shows an individual, representative curve of the ankle torque with values highlighting the beginning, middle, and end of the acceleration period respectively for one participant. This figure shows the increase in the magnitude of the ankle plantarflexor torque through the acceleration and is representative of both conditions. The correlation between step number and ankle plantarflexor torque for the three phases in A1 and A2 are presented in Table 5. For the pre-acceleration, acceleration, and post-acceleration are 0.021, 0.958, and 0.206 respectively, with the acceleration period being significant ($p < 0.05$). For A2, the step correlations were calculated to be -0.226, 0.968, and 0.285 for the pre-acceleration, acceleration, and post-acceleration period respectively with the acceleration period being significant ($p < 0.05$). This shows that there is a strong, direct relationship between step number and ankle plantarflexor torque through the acceleration period.

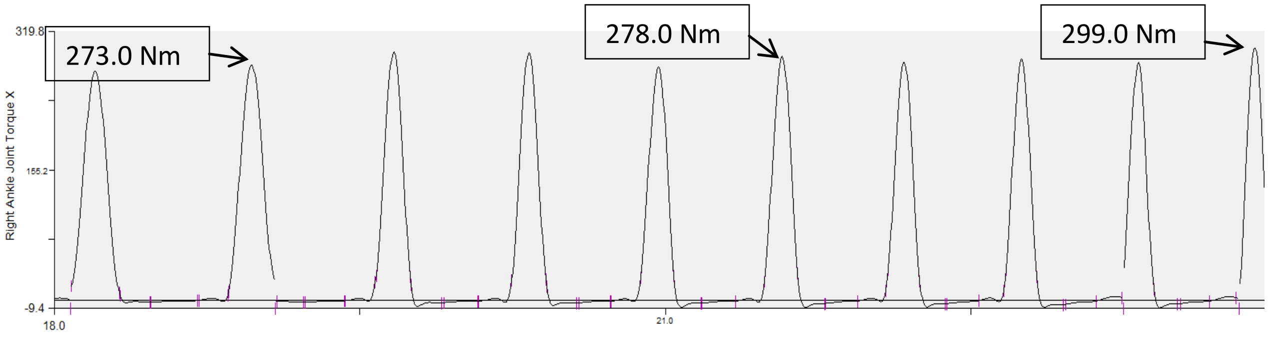

Figure 26: Representation of right leg ankle joint torque during acceleration phase (S3.C1.T1)

Condition	Pre Accel	Accel	Post Accel
A1	0.021	**0.958**	0.206
A2	-0.226	**0.968***	0.285

Table 5: A1 and A2 Ankle plantarflexor torque correlation coefficients between maximum stance phase torque and step number during the acceleration phase; * $p<0.05$

Figure 27 shows the mean peak values of all participants of the three phases for A1. A linear regression beta weight of best fit was calculated for the acceleration period, y = 1.6505x, with $R^2 = 0.9172$ ($p < 0.05$) indicating a significant increase in the magnitude of joint torque at the ankle through acceleration. The pre-acceleration and post-acceleration regression beta weights for A1 were calculated to be y = 0.005x, $R^2 = 0.0005$ and y = 0.0621x, $R^2 = 0.0426$ respectively, indicating no change during the constant states. Figure 28 shows the mean peak values for ankle plantarflexor joint torque in A2. A linear regression beta weight of best fit was calculated for the acceleration period, y = 3.909x and $R^2 = 0.9364$ ($p < 0.05$) also indicative of a significant increase in the acceleration period. The pre-acceleration and post-acceleration regression beta weights were calculated to be y = -0.0406x, $R^2 = 0.0511$ and y = 0.0437x, $R^2 = 0.0813$ respectively, also indicating no change during the constant state periods in ankle torque.

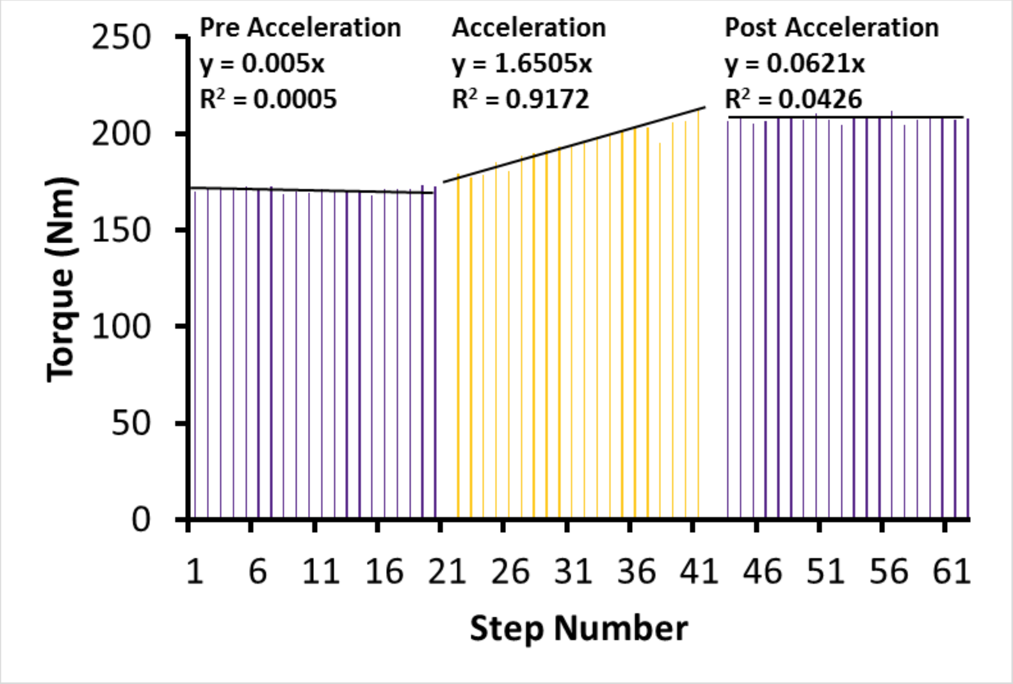

Figure 27: A1 Peak ankle plantarflexor joint torque during pre-, post- acceleration phases

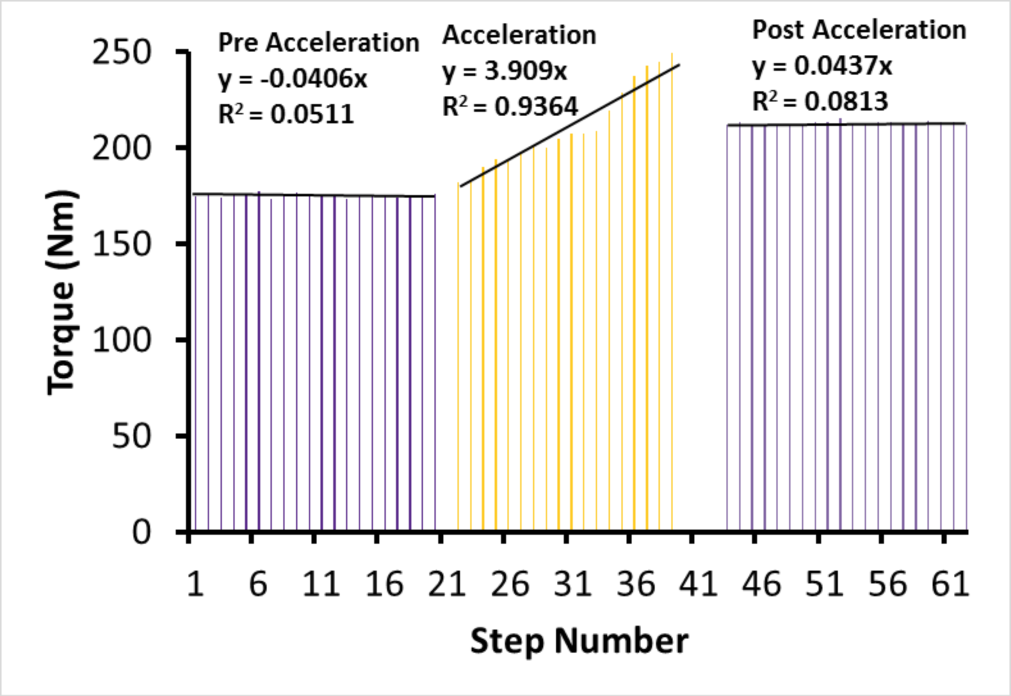

Figure 28: A2 Peak ankle plantarflexor joint torque during pre-, post- acceleration phases

VII). *Ankle Joint Powers*

Figure 29 shows an individual, representative curve of the ankle power with values highlighting the beginning, middle, and end of the acceleration period respectively for one subject. This figure shows the increase in the magnitude of the ankle power through the acceleration and is representative of both conditions. The correlation between step number and ankle joint power for the three phases in A1 and A2 are presented in Table 6. For the pre-acceleration, acceleration, and post-acceleration and they are -0.195, 0.985, and 0.303 respectively, with the acceleration period being significant ($p < 0.05$). For A2, the step correlations were calculated to be -0.430, 0.981, and -0.605 for the pre-acceleration, acceleration, and post-acceleration period respectively, with the acceleration and post-acceleration periods being significant ($p < 0.05$). This shows that there is a strong, direct relationship between step number and torque through the acceleration period. In condition A2, there was a significant inverse relationship during the post-acceleration period.

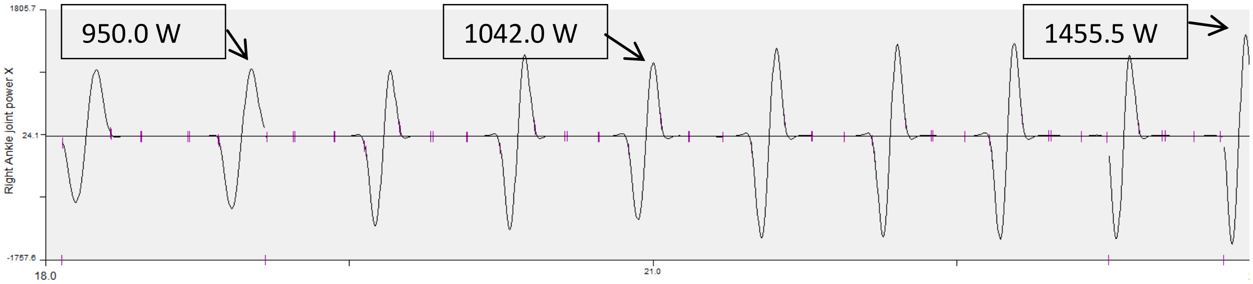

Figure 29: Representation of right leg ankle joint power during acceleration phase (S3.C1.T1)

Condition	Pre-Acceleration	Accel	Post-Acceleration
A1	-0.195	**0.985***	0.303
A2	-0.430	**0.981***	**-0.605***

Table 6: A1 and A2 Ankle power correlation coefficients between maximum stance phase torque and step number during the acceleration phase; * $p<0.05$

Figure 30 shows the mean peak values of all participants of the three phases for A1. A linear regression weight of best fit was calculated for the acceleration period, $y = 26.195x$ with an $R^2 = 0.9707$ ($p < 0.05$) indicating a significant increase in power magnitude through acceleration. For the pre-acceleration and post-acceleration linear regression beta weights were calculated as $y = -0.4014x$, $R^2 = 0.0379$ and $y = -1.1344x$, $R^2 = 0.0919$ respectively, indicating no change in the magnitude of the ankle joint powers during constant state periods. Figure 31 shows the mean peak values for knee joint torque in A2. A linear regression beta weight of best fit was calculated for the acceleration period, $y = 40.029x$ and $R^2 = 0.9622$ ($p < 0.05$) showing a significant increase in ankle joint power during acceleration. For the pre-acceleration and post-acceleration, the regression beta weights were calculated to be $y = -0.6739x$, $r^2 = 0.1848$ and

$y = -0.9729x$, $R^2 = 0.3655$ respectively, again indicating no change in magnitude of the ankle joint powers through the constant state periods.

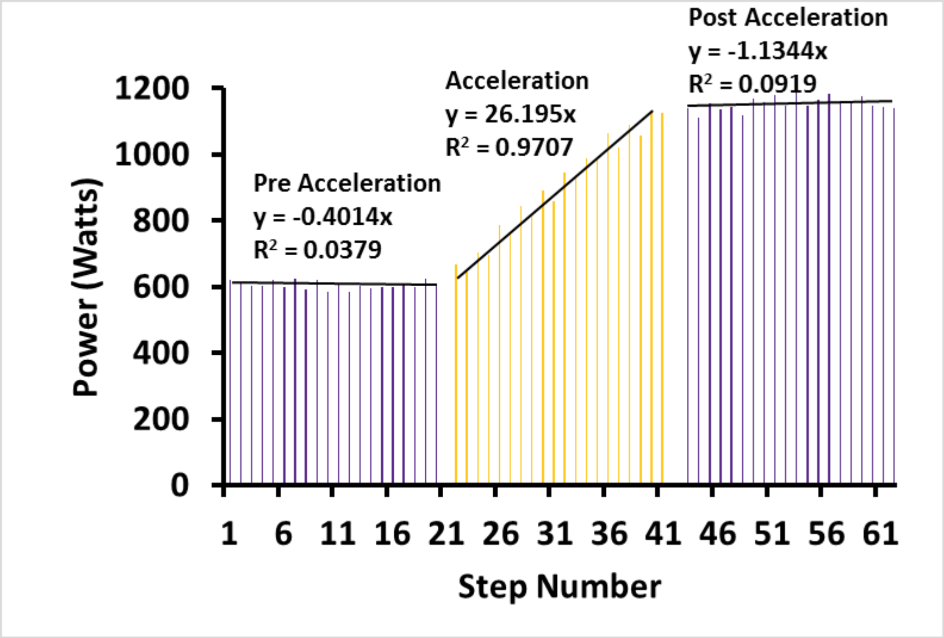

Figure 30: A1 Peak mean sagittal plane, concentric ankle power during pre-, post- acceleration phases

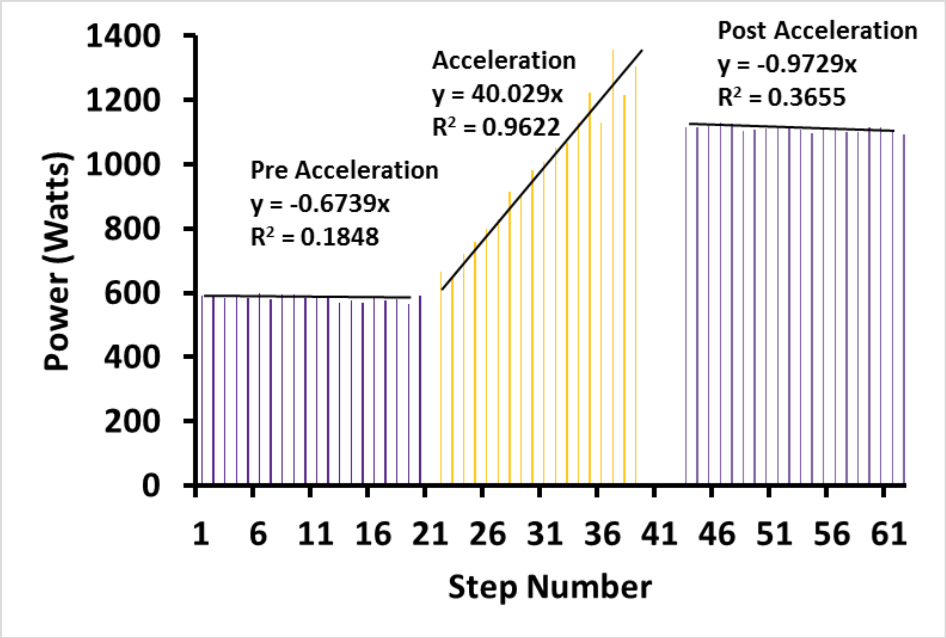

Figure 31: A2 Peak mean ankle power during pre-, post- acceleration phases

VIII). *Regression and Confidence Interval Analysis*

Figures 32a and 33a show schematics of the mean regression beta weights for conditions A1 and A2, respectively, for the joint torques and joint powers. Figure 19a shows the hip joint torque has the greatest beta weight in condition A1 followed by the ankle and then the knee ($p < 0.05$). Figure 20a shows ankle and hip joint torque beta weights did not differ in condition A2 but were significantly greater than the knee ($p < 0.05$). In figures 33b and 33b for the joint powers, it can be seen that the ankle joint power was significantly greater than the hip joint power which was significantly greater than the knee joint powers. Tables 7 and 8 show the mean beta weights from the regression analysis with the minimum and maximum 95% confidence intervals. The results indicate that the beta weights for knee joint torque, ankle joint torque, hip joint power, knee joint power, and ankle joint power were all significantly greater in condition A2 when compared to condition A1 ($p < 0.05$). The hip joint torque beta weights were not significantly different conditions A1 and A2.

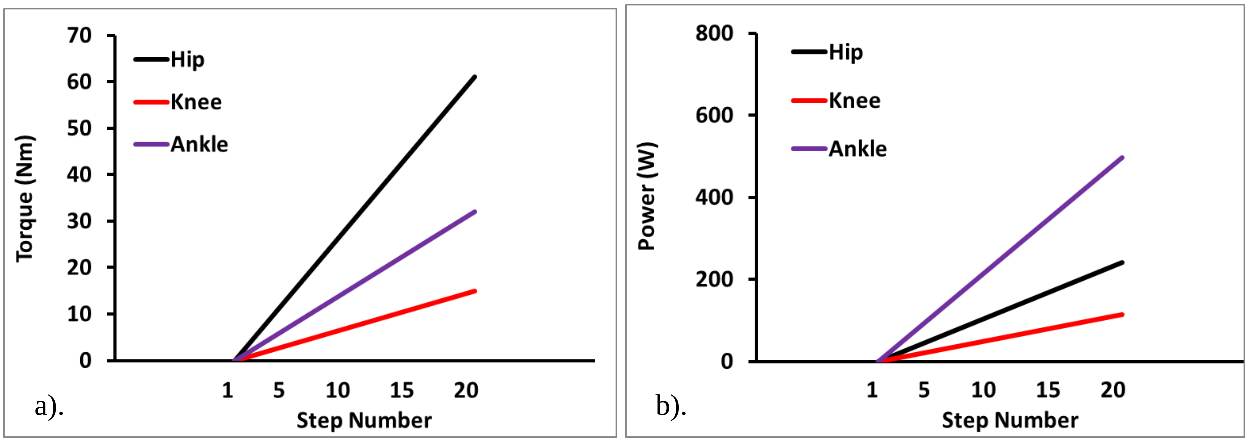

Figure 32 a, b: Depiction of A1 beta weight slopes/ rate of change shown over the sequence of steps during the acceleration phase. Value at first step represents initial value during acceleration phase for each joint and higher values show the amount of change from the initial value.

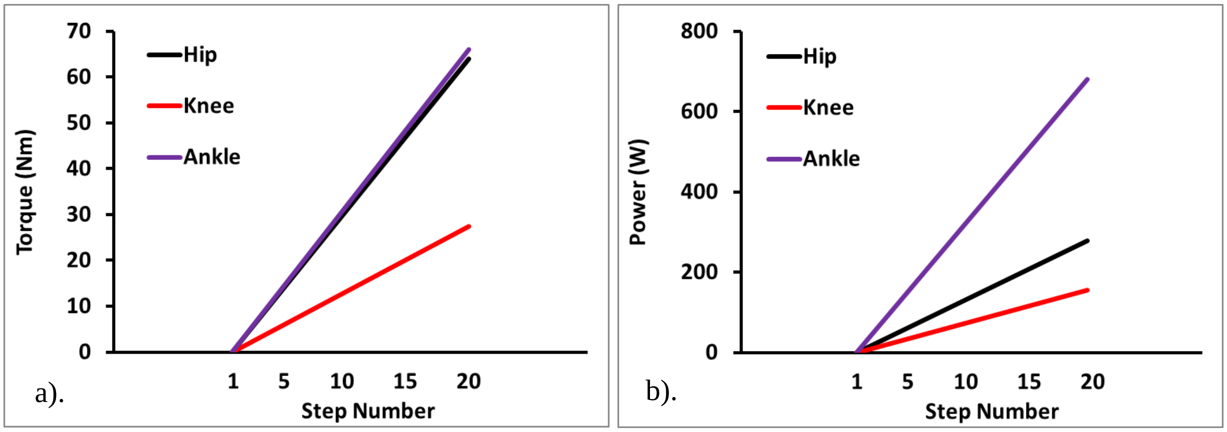

Figure 33 a, b: Depiction of A2 beta weight slopes/ rate of change shown over the sequence of steps during the acceleration phase. Value at first step represents initial value during acceleration phase for each joint and higher values show the amount of change from the initial value.

Condition	**Hip Torque**			**Knee Torque, Φ**			**Ankle Torque, Φ**		
	-95% C.I.	*Mean*	*+95% C.I*	*-95% C.I.*	*Mean*	*+95% C.I*	*-95% C.I.*	*Mean*	*+95% C.I*
A1, α, β	3.14	3.23	3.32	0.71	0.81	0.90	1.49	1.65	1.72
A2, Ω	3.50	3.80	4.10	1.04	1.16	1.28	3.69	3.91	4.13

Table 7: Beta weights from regression analysis and 95% confidence intervals for hip, knee, and ankle joint torques. α: hip > ankle, $p < 0.05$; β: ankle > knee, $p<0.05$; Ω:hip & ankle > knee, $p<0.05$; Φ: condition A2 > A1, $p<0.05$

Condition	**Hip Power, Ψ**			**Knee Power, Ψ**			**Ankle Power, Ψ**		
	-95% C.I.	*Mean*	*+95% C.I*	*-95% C.I.*	*Mean*	*+95% C.I*	*-95% C.I.*	*Mean*	*+95% C.I*
A1, γ, δ	12.3	12.9	13.3	5.67	6.06	6.44	25.1	26.2	27.3
A2, ε, θ	14.9	16.5	17.9	8.19	9.16	10.1	38.3	40.0	41.8

Table 8: Beta Weights from regression analysis and 95% confidence intervals for hip, knee, and ankle joint powers.

γ: ankle > hip, $p < 0.05$; δ: hip > knee, $p<0.05$; ε: ankle > hip, $p<0.05$; θ: hip > knee, $p<0.05$; Ψ: condition A2 > A1, $p<0.05$

IX). *Summary*

Based on these results, it can be concluded that there is a linear increase in the magnitudes of the joint torques and joint powers at the hip, knee, and ankle during acceleration while running. Based on the calculated regression beta weights and magnitudes of the rates of change at the three joints it can also be concluded that in both conditions that the hip joint and ankle joint are the main contributors to acceleration during running. The hip joint torque, in condition A1, had the greatest rate of change magnitude as seen in figure 32 and table 7 being significantly greater than the ankle joint, which was significantly greater than the knee joint ($p<0.05$). The hip and ankle torque beta weights did not differ significantly in A2 but were significantly greater when compared to the knee ($p<0.05$). The ankle joint power had the greatest rate of change magnitude in both conditions, as seen in figures 32 and 33 and table 8 and based on the beta weights an emphasis should be placed on the change in power magnitude at the ankle. The pre-acceleration and post-acceleration constant state correlations and beta weights agree with the findings of previous studies in that there were no changes observed in the magnitude of joint torques and joint powers, except in the pre-acceleration hip joint power in A1 and the post-acceleration ankle power in A2.

CHAPTER 5: DISCUSSION

The purpose of this study was to quantify lower extremity joint torques and powers during constant speed running and during running while accelerating at two rates of acceleration (0.40 ms^{-2} and 0.80 ms^{-2}) between a baseline velocity of 2.50 ms^{-1} to 6.00 ms^{-1}. It was hypothesized that lower extremity, sagittal plane joint torques and joint powers would positively and linearly increase throughout the acceleration phase of running which was found to be supported based on the results of this study. This section will further investigate the findings of this study and will be broken down into the following sections: 1).Comparison to the Previous Literature on Running Velocity, 2). How Humans Accelerate When Running And Comparison Of Accelerated Running Conditions, 3). Applications of Present Study Results, 4). Limitations of Present Study, and 5). Conclusion.

Comparison to the Previous Literature on Running Velocity

The joint kinetics of running gait have been well documented in the field of biomechanics, starting with the classic study by Winter (1983). The joint torque curves found in this study aligned with those found in the results of Winter (1983) in both their magnitude of the calculated joint torques and joint powers and general figure shape when matched for velocity. The average velocity in Winter's (1983) was 2.72 ms^{-1}. The peak hip, knee and ankle joint torques during the stance phase were approximately 80 Nm, 160 Nm, and 180 Nm respectively. When compared to the results of this study at approximately the same velocity the hip, knee, to a lesser degree, and ankle joint torque curve peak values are approximately the same, being 75 Nm, 110 Nm, and 183 Nm, respectively, with the notion that there is always some variance between participants. The joint power curves also had similar magnitudes and the similar general shape at the hip, knee, and ankle joint. The magnitudes calculated by Winter (1983) for the hip, knee, and ankle joint powers were approximately 100 W, 220 W and 600 W which were approximately the same for the similar velocity in this present studywhich were calculated to be approximately 130 Nm, 190 Nm, and 660 Nm, respectively. The results of this study showed peaks in the hip torque and power curves at the beginning of stance, peaks in the knee torque and power curves in mid-stance, and peaks in the ankle torque and power curves near the end of stance which again aligns with the curves of Winter (1983). This study by Winter (1983) is considered to be the seminal study in quantifying running gait kinetics and has been cited in many running biomechanics research papers.

As stated previously, the majority of the running research relating to velocity has investigated the effect of running velocity on joint kinetics in constant state velocity increments. This research was very important to the formulation of the hypothesis of this study. The previous research showed that when running velocity was increased the magnitude of the joint torques and joint powers directly increased in relation to the velocity (Belli et al., 2002; Dorn et al., 2012; Schache et al., 2011). The findings of these studies all showed the hip and ankle to increase the most in joint torque and joint power magnitude which aligns with the findings of the present study which also showed this based on the regression beta weight analysis (figures 19 and 20 and tables 7 and 8 of the Results section) that the hip and ankle contribute the greatest amount during acceleration. It was also found that the knee joint torques and powers increased during acceleration but not to the same magnitude as the hip and ankle which is similar to the results found in the studies of Belli et al (2012) and Schache et al (2011).

The results of this study are in alignment with the findings of the previous literature investigating increases in velocity in constant state increments when matched relatively to the velocity in the acceleration period of this study. The results of Belli et al (2002) at 4.00 ms^{-1} had joint torques of similar magnitudes at the hip and ankle to the present study as well as joint powers since they were within 5% of the calculated joint torques and joint powers at the hip and ankle. There was a large difference at the knee joint for joint torque (29% less) and joint power (30% less). In the same study, similar joint torques and joint powers were seen with the present study at the hip and ankle joints (within 5%) at 6.00 ms^{-1} with there being a large difference with the present study in the magnitude of the knee joint torque and joint power (55% and 107% greater respectively) (Belli et al., 2002). This is one study that the results of the present study aligned with when matched for velocity with the exception of the knee joint torque and powers at 6.00 ms^{-1}.

The results of the present study also agreed with some results of Schache et al (2011) and Dorn et al (2012) when the joint torques and joint powers were normalized to body mass and matched for approximate velocity. These studies both tested four different running velocities and the two that were within the range of the velocities of this study were 3.50 ms^{-1} and 5.00 ms^{-1}. When matched to 3.50 ms^{-1} our normalized results for hip joint torque were not similar to the results of these studies being 40% greater than Dorn et al (2012) and 28% less than Schache et al (2011) but our hip joint torque results did fall in between the results of these studies. The results for knee joint torque were similar in magnitude to the results of Dorn et al (2012) with the present result being 16% less than the results of Dorn et al (2012). The results of the present study were 45% less than the results of Schache et al (2011) for knee joint torque. The calculated ankle joint torques of the present study were the same in magnitude to the results of Schache et al (2011) having no difference (0%) in magnitude. Our results were only 8% less than the calculated ankle joint torques of Dorn et al (2012). At 5.00 ms^{-1}, similar results were observed in the present and Dorn et al (2012) with the present study results being 22% less than Dorn et al

(2012). The results of Schache et al (2011) were 40% greater than the present study results for hip joint torque. The results for knee joint torque at 5.00 ms^{-1} for the present study were 41% less than the results of Dorn et al (2012) and 59% less than Schache et al (2011). The present study had similar ankle joint torque magnitudes when compared to Schache et al (2011) at 5.00 ms^{-1}, with the presen study's results being 8% less than their results. The ankle joint torque results of the present study were 3% greater than the results of Dorn et al (2012). When comparing joint powers of the present study to the results of Schache et al (2011) the only calculated variable within 25% was ankle joint power for 3.50 ms^{-1} , with the present study results being 22% less than those of Schache et al (2011). The hip joint and knee joint powers for the present study were 40% greater than and 66% less than the results of Schache et al (2011). An overview of these results comparisons can be seen in Table 9 below. While there were some similarities to aforementioned similarities the differences should be considered.

Some of the results for the present were not in agreement with the results of the previous literature investigating increases in running velocity in constant state increments (Belli et al., 2002; Dorn et al., 2012; Schache et al., 2011). The reasons for these differences can only be speculated but one such reason could be the previous literature was performed overground and the present study was done on an instrumented treadmill. There is disagreement in the literature as to whether running on a treadmill is the same as overground kinematically and kinetically (Lee & Hidler, 2008; Riley et al., 2008; Van Caekenberghe et al., 2012). Taking this into account, another reason for the difference could be the differences in vertical and horizontal kinematics and kinetics. The present study used an instrumented treadmill and there is no change horizontal displacement and there could have been an increase in vertical displacement for this reason. There is also normal variation in running gait between participants and this could account for these differences. The present study used runners of all levels from recreational to division I cross country runners whereas the referenced studies all used elite level sprinters which could account for the differences observed especially at the hip joint torque given the increased role of the hip musculature during sprinting. Finally, it could simply be errors in the calculations coming from the 3D position data or the force plate.

	Belli et al (2002)		**Schache et al (2011)**		**Dorn et al (2012)**	
Matched Velocity	4.00 ms^{-1}	6.00 ms^{-1}	3.50 ms^{-1}	5.00 ms^{-1}	3.50 ms^{-1}	5.00 ms^{-1}
Hip Torque	**-5%***	**-11%***	-28%	+40%	+40%	**-22%***
Hip Power	**-5%***	+33%	+55%	+40%	N/A	N/A
Knee Torque	-29%	-55%	-45%	-59%	**-16%***	-41%
Knee Power	-30%	-107%	-55%	-66%	N/A	N/A
Ankle Torque	**+1.5%***	**+3%***	**0%***	**-8%***	**-8%***	**+3%***
Ankle Power	**+5%***	**+11%***	**-14%***	-32%	N/A	N/A

Table 9: Comparison of present study results to previous literature. + indicates present study results were greater than the results of corresponding study results, - indicates present study results were less than corresponding study results. *percentages within ± 25% were considered similar.

There have been few running studies that have investigated accelerated running in humans. The study by Van Caekenberghe et al (2013) is the one of the only investigations on human acceleration. However, the findings of Van Caekenberghe et al (2013) found that there was no increase in the joint torque magnitude which is the opposite of the findings of the present study which found there to be a linear increase in the joint torques in the hip, knee, and ankle with the hip and ankle being the primary contributors based on the regression analysis discussed in the results section (Figures 19a and 20a). Van Caekenberghe et al (2013) suggest that the reason we accelerate and have a greater propulsive ground reaction is due to the body orientation change. They state that the muscular contributions of the joints are the same but because of there is a more forward lean in terms of body orientation this results in a change in the direction of the ground reaction forces which is how we accelerate . The present study did not take into account body lean but again did find there to be a significant linear increase in joint torque magnitude during the acceleration period on the instrumented treadmill. Van Caekenberghe et al (2013) did find that the joint powers did increase (did not quantify this increase in a regression) and this observation is in alignment with the findings of the present study which found a linear increase in the joint powers at the hip, knee, and ankle. Our results can be explained by an increase in the magnitude of the joint torques which is one of the factors of joint power whereas Van Caekenberghe et al (2013) attribute their increase to an increase in the joint angular velocity. Angular velocity was not a recorded variable in the present study. The findings of the present study do not align with one of the few human accelerated running studies but they do align with those done in animal models.

McGowan et al (2004) performed a study using tammar wallabies and they found that the hip and knee joint torques increased during acceleration with the ankle not increasing. This aligns with the results of the present with the exception being we did observe an increase in the ankle joint. The reason for this difference could be explained by the difference in the anatomy of the ankle joint in a tammar wallaby. Using turkeys, another study found that during two different accelerations (moderate and fast) an increase in the magnitude of the joint torques and joint powers was observed in both acceleration conditions when compared to the steady state which is similar to the results of the present study (Roberts & Scales, 2004).

Overall, the findings of this study do align with the results of the previous literature- when matched approximately for velocity- that investigated the changes in joint kinetics when running velocity was increased in constant state increments. The findings of the present study do not align with the results of Van Caekenberghe et al (2013) as increase was observed during the acceleration period in the magnitude of the joint torques at the hip, knee, and ankle. The results for joint powers in the same study did align with that of the present study in that both suggest there to be an increase during acceleration. The results of the present study were similar to the findings of Roberts and Scales (2004) and McGowan et al (2004) which both found there to be increases in the joint torques and joint powers during acceleration.

How Humans Accelerate When Running And Comparison Of Accelerated Running Conditions

Based on the results of this study the general running kinematics (as explained in the literature review) remain the same when humans start to accelerate but the joint kinetics of the lower extremity are what change. For this study, correlation and a regression analyses were used to determine the strength of the relationships between joint torques and step number and the magnitude of the increase in the joint torques and joint powers during the acceleration phase at the hip, knee, and ankle joint. As stated, there is a linear increase in the joint torques and joint powers at the hip, knee, and ankle when humans accelerate. However, the contributions based on the rates of change per step of the hip, knee, and ankle are not equal as seen by the beta weight regression analysis and the confidence intervals for the joint torques and joint powers. The differences between conditions A1, acceleration rate of 0.40 ms^{-2},and A2, acceleration rate of 0.80 ms^{-2}, for beta weights were also analyzed.

In condition A1, the beta weights for the joint torques revealed that the hip has a significantly greater muscular contribution than the next greatest joint torque beta weight during the acceleration having a beta weight of 3.23. The 95% confidence interval for the hip joint torque beta weight was 3.14 to 3.32 ($p < 0.05$). The ankle joint for condition A1 had the next greatest beta weight with a mean beta weight of 1.65, which was significantly less than the hip joint torque ($p < 0.05$). The 95% confidence interval for the ankle joint torque was 1.49 to 1.72. Finally, the knee joint torque does increase during acceleration but contributes the least during acceleration based on the regression analysis of the beta weights. The mean joint torque beta weight was calculated to be 0.81, which was significantly less than the ankle joint ($p < 0.05$). The 95% confidence interval for the knee joint torque was 0.71 to 0.90. These were the findings of condition A1 which was the slower rate of acceleration. These findings are similar to that of A2 with one difference.

In condition A2, the beta weights revealed that the hip joint torque and ankle joint torque were not significantly different from one another, unlike condition A1 where they were significantly different. The mean beta weight for the hip joint torque was 3.80 and 3.91 for the ankle joint torque which were not significantly different. The 95% confidence interval for the hip joint torque was 3.50 to 4.10 and 3.69 to 4.13 for the ankle joint torque. The hip joint torque beta weight in condition A2 was not significantly different from the hip joint torque beta weight in A1 but the ankle joint torque beta weight was significantly different from A1. The hip and ankle joint torque beta weights were significantly greater than the knee joint torque beta which had a mean beta weight of 1.16 ($p < 0.05$). The 95 % confidence interval for the knee joint torque beta weight was 1.04 to 1.28. This result is similar to that of condition A1 in that the knee joint contributes the least when we accelerate when running. When comparing the results between condition A1 and A2 the hip joint torque beta weights are not significantly different but the knee torque and ankle torque beta weights are significantly different indicating a difference in the slow versus fast rates of acceleration. When we analyze the beta weights for the hip, knee, and ankle joint powers we see a different trend.

The order of peak magnitudes for joint powers is the same for both conditions A1 and A2 but is not the same when compared to the joint torques. The beta weight regression analysis revealed that the ankle joint had the greatest increase in peak, power per step during the acceleration period having a mean beta weight of 26.2 and 40.0 for A1 and A2 respectively which was significantly greater than the hip joint power and knee joint power ($p < 0.05$). The 95% confidence intervals for the ankle joint power were 25.1 to 27.3 and 38.3 to 41.8 for A1 and A2, respectively. This also denotes that there was a significant difference in ankle joint power between the slow and fast rates of acceleration (conditions A1 and A2). The hip joint power beta weights for both conditions A1 and A2 were significantly less than the ankle but were significantly greater than the knee joint powers ($p< 0.05$). The mean hip joint power beta weights for A1 and A2 were 12.9 and 16.5, respectively and the 95% confidence interval were 12.3 to 13.3, and 14.9 and 17.9, respectively. This also indicates that there was a significant difference between conditions A1 and A2 ($p < 0.05$) with the larger acceleration having higher per step rate of change in hip power. Finally, the knee joint powers were significantly less than both the ankle joint and hip joint powers. The mean knee joint beta weights for conditions A1 and A2 were 6.06 and 9.16, respectively and the 95% confidence intervals were 5.67 to 6.44 and 8.19 to 10.1, respectively. This is again indicative of a difference between conditions A1 and A2 for the knee joint powers ($p < 0.05$).

In summary, the results of this study indicate that when humans accelerate the muscular contributions, made evident by the joint torque beta weights, indicate that in both conditions A1 and A2 the hip and ankle joint torques have the greatest contribution during acceleration and increase the greatest in magnitude per step. This could be due to the role the hip has in increasing the step rate to move the limbs faster and the ankle in increasing step length (Dorn et al., 2012; Fukunaga et al., 1980). In both conditions it was also apparent the knee joint torque beta weight was the lowest in magnitude and therefore the knee joint contributes the least to acceleration in both conditions A1 and A2 and increases the least per step. The ankle joint torque had different results between the two conditions. In condition A1 it was significantly less than the hip joint torque beta weight but was not significantly different in condition A2. This suggests again that the hip and ankle joint musculature are the primary contributors to the acceleration phase of running based on the rates of change in these joint for joint torque. The joint powers had the same results in both conditions where the ankle had the greatest joint power beta weight magnitude, followed by the hip joint and lastly by the knee joint. This suggests that the muscles around the ankle joint are performing a more powerful concentric contraction than the hip and knee joint musculature and increasing the most per step followed by the hip joint increasing the second most per step, and lastly the knee joint. The joint powers differed significantly at all three joints between conditions A1 and A2 with condition A2 having significantly greater joint power beta weights. This indicates that there are more forceful concentric contractions at all three joints in the lower extremity when humans accelerate at a faster rate. Overall based on the results of joint kinetics it could be said the ankle is the primary joint driving acceleration during running.

Applications of Present Study Results

Running is an integral part of many sports such as track, football, soccer, basketball, etc. In all of these sports, there are periods where changes in velocity occur for a variety of reasons. Based on the protocol, the results of the present study could be most applicable to the way in which sprinters accelerate on a track. Sprinters experience a gradual increase in running velocity through gradual acceleration before the rate of acceleration is maximized. In the present study, the particpants performed a a gradual, but constant rate of accelerationg for approximately six to nine seconds depending on the rate of acceleration being tested. It could be argued that this is a similar time frame for acceleration when sprinting on a track. Acceleration in the other sports mentioned may usually occur in a very short span, taking 3 or 4 accelerating steps perhaps. This makes applying the findings of the present study difficult given the longer period of accelerated running the participants performed

However, it must also be considered that during sprinting acceleration on a track the rate of acceleration may not be constant as it was in the present study’s protocol. This could suggest acceleration being a skill as some track athletes could be superior to other athletes at accelerating more efficiently and constantly. Manipulating the rate of acceleration was important to the present study to insure that all participants did accelerate similarly and took the same, relative number of steps. Quantifying the rates of acceleration for track athletes during a sprinting event (100 m or 200 m) is a possible future direction for the research on acceleration based off of this idea of acceleration being a skill. Coaches could emphasize the movements of the ankle first followed by the hip to help in the improved performance of track athletes out of the blocks and during the acceleration period during an event.

Limitations of Present Study

This study is not without some limitations. First, there is some error that occurs during the testing protocol by the motion capture cameras that is result of residual error or movement artifact of the reflective markers. This in turn results in small discrepancies in the inverse dynamic calculations. Another limitation of this study is that a Bertec instrumented treadmill was used for the data collections and there is disagreement in the literature as to whether treadmill running is the same as over ground running (Lee & Hidler, 2008; Riley et al., 2008; Van Caekenberghe et al., 2012). A treadmill was used to be able to collect multiple steps and have the rates of acceleration be the same in each condition and between trials. When entered into the spreadsheet, some of the treadmill data were deleted due to large discrepancies or measurement errors but this amounted to only approximately 1.5% of the total steps collected and entered.

The study also had a small sample size of 15 healthy, young runners. Within this sample there were 8 females and 7 males and each had various degrees of running experience. Some were division I cross country runners with coaching and others were purely recreational. There were natural differences in the running gait of each participant which could have affected the results to some capacity. One way in which this affected the results is not all the particpants took the same number of steps during the acceleration- particularly in condition A2. The participants that took more steps happened to be the less heavy particpants and this resulted in there being a decrease in the mean torque and power values for the greater step numbers for some variables which then affected the regression beta weights. This would be difficult to control for but should mentioned as a limitation. Another potential limitation is the comfort level of the participants on the instrumented treadmill. It is somewhat different from a normal treadmill as it has two belts moving simultaneously with a small space between the two belts. Depending on how the participants ran on the treadmill their stance width could have been slightly wider if they ran on both belts or slightly more narrow if they were only on one belt.

Conclusion

It was hypothesized that lower extremity, sagittal plane joint torques and joint powers would positively and linearly increase throughout the acceleration phase of running. Based on the results of this study, the hypothesis was supported. There were some differences in the amount each joint contributes to acceleration during running but the general finding was that the hip and ankle contribute the greatest amount to acceleration during running based on the idea that the rates of change in joint torque and joint power magnitude were greatest at these two joints. It is suggested that the ankle is primary joint contributing to accelerated running based on the very high increases in magntidue per step for joint power. The knee joint also contributes but not to the same degree as the hip and ankle and this again can be seen by the results for therate of change beta weights for the joint torques and joint powers for the hip, knee, and ankle. It was also observed that magnitude of the increase per step was different between the two conditions for the knee torque, ankle torque as well as the hip, knee, and ankle joint powers with condition A2 having a greater increase in magnitude per step when humans accelerate at a faster rate. This study refutes the findings of Van Caekenberghe et al (2013) and therefore suggests that further research is needed to add to the minimal research investigating the acceleration phase of running.

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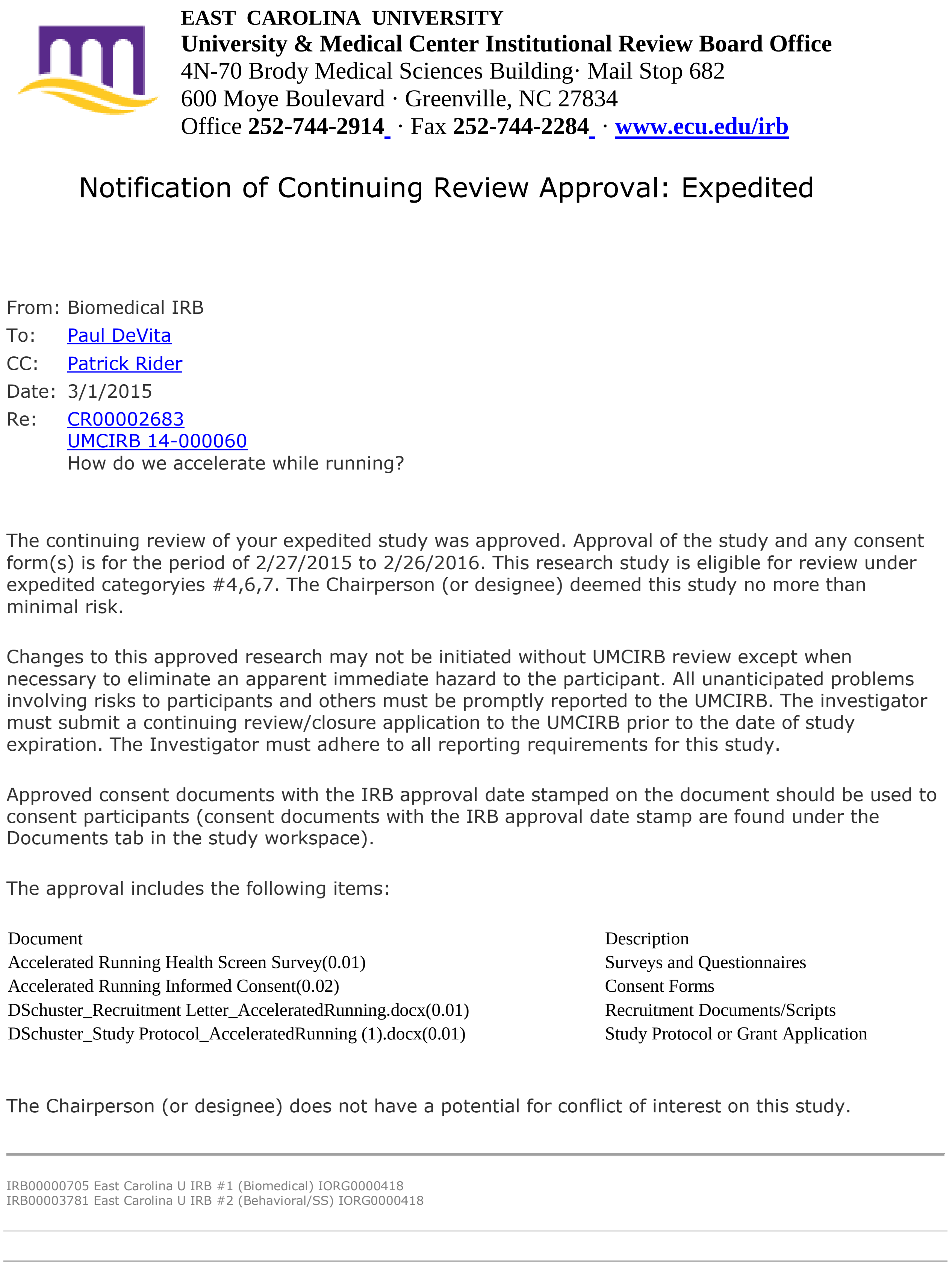
APPENDIX A: IRB Approval Letter



EAST CAROLINA UNIVERSITY
University & Medical Center Institutional Review Board Office
4N-70 Brody Medical Sciences Building· Mail Stop 682
600 Moye Boulevard · Greenville, NC 27834
Office **252-744-2914** · Fax **252-744-2284** · **www.ecu.edu/irb**

Notification of Continuing Review Approval: Expedited

From: Biomedical IRB
To: Paul DeVita
CC: Patrick Rider
Date: 3/1/2015
Re: CR00002683
UMCIRB 14-000060
How do we accelerate while running?

The continuing review of your expedited study was approved. Approval of the study and any consent form(s) is for the period of 2/27/2015 to 2/26/2016. This research study is eligible for review under expedited categoryies #4,6,7. The Chairperson (or designee) deemed this study no more than minimal risk.

Changes to this approved research may not be initiated without UMCIRB review except when necessary to eliminate an apparent immediate hazard to the participant. All unanticipated problems involving risks to participants and others must be promptly reported to the UMCIRB. The investigator must submit a continuing review/closure application to the UMCIRB prior to the date of study expiration. The Investigator must adhere to all reporting requirements for this study.

Approved consent documents with the IRB approval date stamped on the document should be used to consent participants (consent documents with the IRB approval date stamp are found under the Documents tab in the study workspace).

The approval includes the following items:

Document	Description
Accelerated Running Health Screen Survey(0.01)	Surveys and Questionnaires
Accelerated Running Informed Consent(0.02)	Consent Forms
DSchuster_Recruitment Letter_AcceleratedRunning.docx(0.01)	Recruitment Documents/Scripts
DSchuster_Study Protocol_AcceleratedRunning (1).docx(0.01)	Study Protocol or Grant Application

The Chairperson (or designee) does not have a potential for conflict of interest on this study.

IRB00000705 East Carolina U IRB #1 (Biomedical) IORG0000418
IRB00003781 East Carolina U IRB #2 (Behavioral/SS) IORG0000418

APPENDIX B: Treadmill Belt Velocity Data

The figures below show the treadmill velocity data in relation to the horizontal ground reaction forces and the minimal fluctuation that occurs in the belt velocity when the participants pushed off and made initial contact. The treadmill belt velocity had a coefficient of variation between 1 and 1.5 for the two velocities test (2.5 ms^{-1} and 4.5 ms^{-1}) for both the light mass and heavy mass participants. In figures 34-37 when the partipants made initial contact with the treadmill belt there is a slight decrease in the velocity of the belt and then conversely a slight increase when the participants push off at the end of the stance phase. This is what typically happens in normal running overground so it may aid in the argument of treadmill running being kinetically similar to overground running.

However it should also be taken into account that some of the force and energy exterted into the belt is translated into the motor of the treadmill and is therefore lost from the data. This could potentially affect the results of the present study to a certain degree. This notion is a minor concern for the results validity.

This data was not included in the results as it was only collected on two participants who had varying masses and heights as well as being one male and one female.

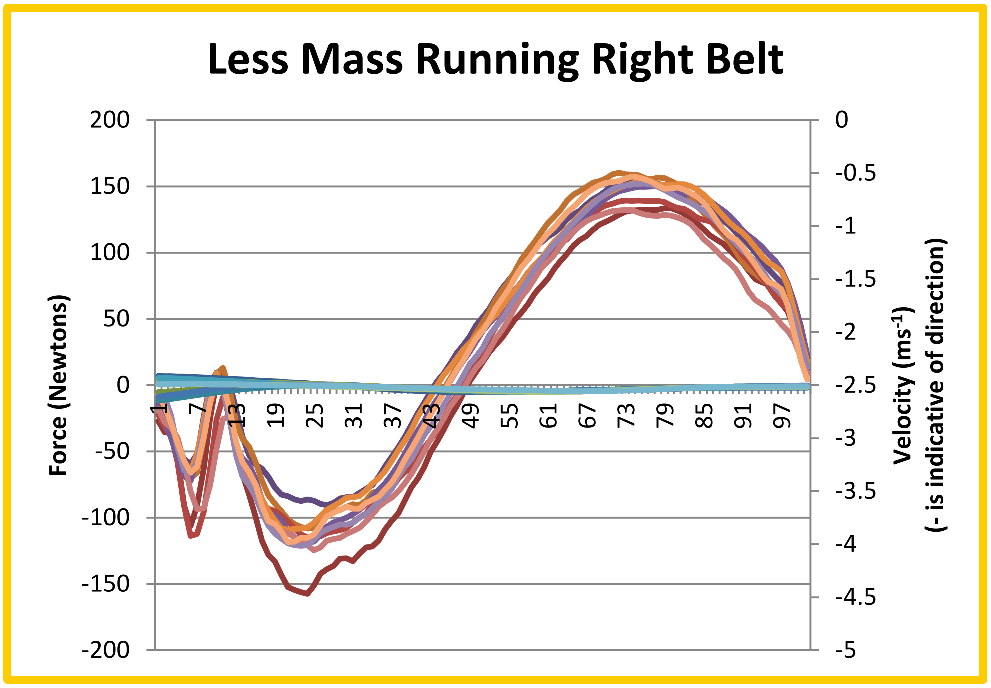


Figure 34: Light Mass 2.5 ms^{-1} running velocity versus horizontal GRF

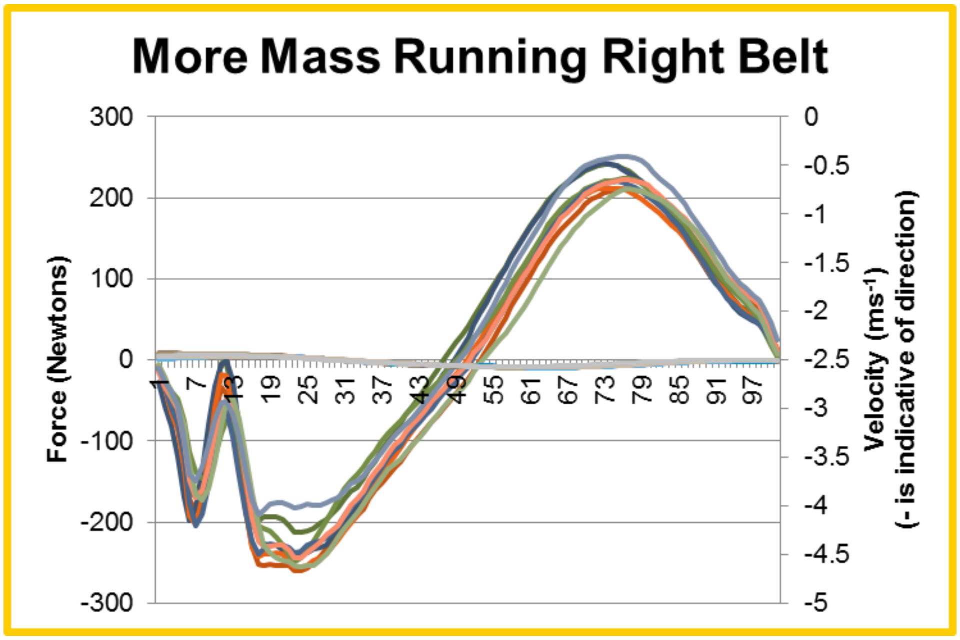


Figure 34: Heavy Mass 2.5 ms^{-1} running velocity versus horizontal GRF

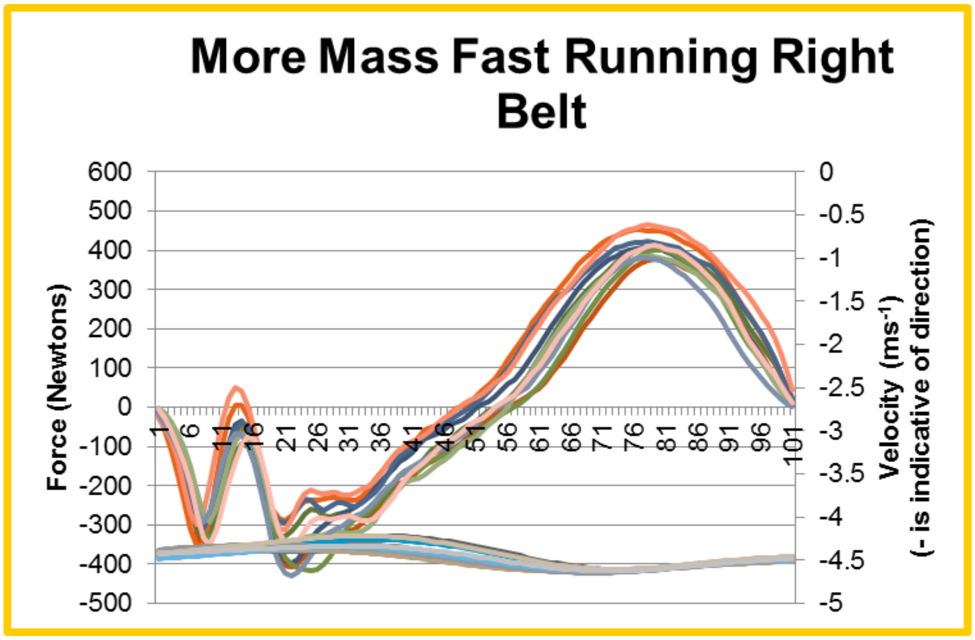


Figure 34: Less Mass 4.5 ms^{-1} running velocity versus horizontal GRF

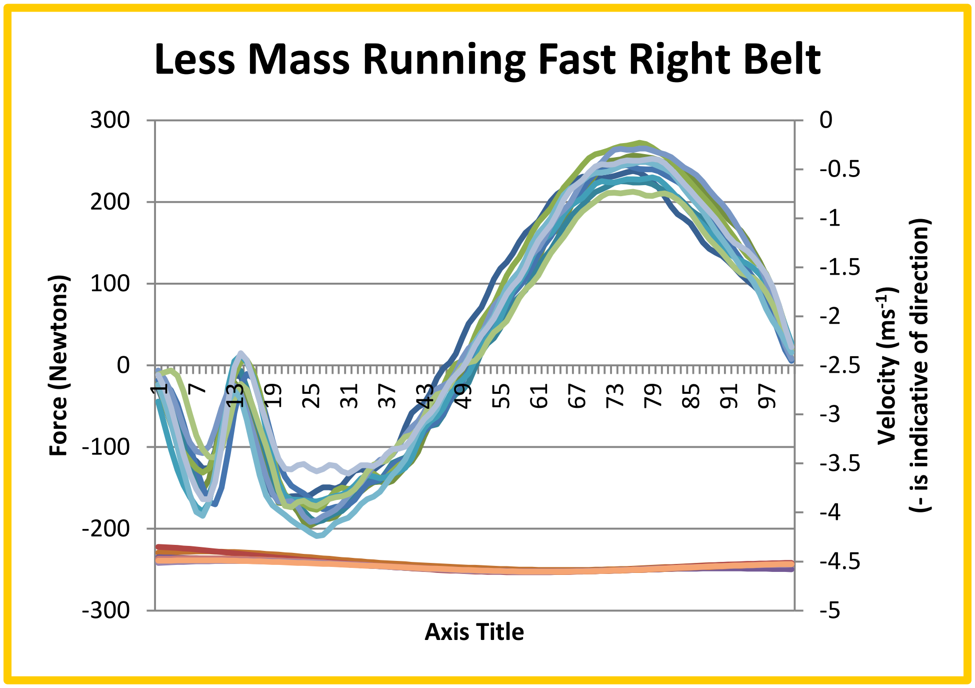


Figure 34: Heavy Mass 4.5 ms^{-1} running velocity versus horizontal GRF